Short and long-term behaviour of dental cast restorations under compressive stresses

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DECLARATION

This thesis is the original work of the author, with the exception of the help and guidance from the individuals acknowledged in the text.

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Abstract

The cement which retains a crown on its preparation is essential. The literature revealed factors which might influence the performance of cast restorations in the oral environment with the commonest cause of failure of crowns having been reported to be recurrent dental caries. It appeared probable that damage or loss of the cement film was a precursor to the development of dental decay around a crown. However, the review of the literature showed that there was a lack of information about the behaviour of the cement film which luted indirect restorations.

This study examined indirectly the behaviour of zinc phosphate cement used to lute cast gold crowns. In order to standardise an *in vitro* method, cast nickel chrome dies were fabricated to represent natural molar teeth prepared to receive full gold crowns, whilst the cast gold crowns were produced using standardised laboratory techniques. The crowns were subjected to direct axial loading using an Instron Universal Testing Machine. In the first series of experiments these loads were static and in the second series, dynamic. The response to loading was monitored using two uni-axial miniature strain gauges; these were located on opposing external surfaces of the crowns about 1 millimetre coronal to the margins.

Initial work was undertaken to develop methods to control the distribution and film thickness of the cement beneath the crown. These were investigated together with the effects of preparation geometry and casting rigidity. Cement

coverage had a strong influence on strain distribution. Increasing the cement film thickness led to an increase in compressive microstrain in the walls of the crowns, whilst both increasing the convergence angle and reducing the axial wall height of the preparation increased the axial compressive microstrain. However, the influence of axial wall height was greater than that of convergence angle. The influence of the casting rigidity showed an interesting finding Heat treatment of the crown did not show differences in microstrain compared with the cast control group, but increasing the occlusal surface thickness by a factor of three doubled the compressive axial microstrain recorded.

The strain response curves for uncemented, partially cemented and fully cemented gold crowns, to increasing static loads were recorded: these data were used to characterise the microstrain measurements from the later series of dynamic loading experiments.

In the dynamic loading series, the crowns were partially cemented on their dies and axial loading was applied between 0-450N for approximately 300,000 cycles with a frequency of about 2Hz. In a series of four experiments, pairs of crowns were tested dry, immersed in water, in water and acid and in acid alone (pH 2.74). The results were plotted as microstrain against the number of cycles. These showed that hydration had a strong effect on the strain recorded on the axial surfaces of the cemented crowns compared with the dry samples. Whilst the presence of pure water decreased the microstrain, the presence of acid increased it. The pronounced effects were recorded in the presence of acidic media.

Microscopic investigation using scanning electron microscope was carried out. The aim was to correlate the measurements from the strain gauges with the microscopic appearance of the cement film after sectioning the crown-die assemblies longitudinally through the centre of the strain gauges. The photomicrographs showed damage to the luting cement was increased by the presence of moisture and further increased by the presence of acid. There was however no apparent correlation between the strain gauge readings and the microscopic views.

The conclusions drawn from this study were that the use of axially-placed strain gauges provided a method for monitoring the strain in the axial walls of cast gold crowns cemented with zinc phosphate cement. The strains were indicative of the condition of the bond between the die and crown provided by the cement. This method has further application in the non-destructive monitoring of the bond provided by luting cements between an indirect restoration and the underlying preparation.

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Nomenclature

No.	Symbol	Definition
1	E_{crown}	Modulus of elasticity of gold crowns.
2	E_{die}	Modulus of elasticity of nickel chromium dies.
3	G_{cem}	Modulus of shear of zinc phosphate cement
4	t_{cem} =	Thickness of zinc phosphate cement.
5	t_{crown}	Thickness of gold crown.
6	t_{die}	Thickness of the nickel chromium die.
7	l	Distance of the centre of strain gauge from the top of the crown.
8	F	Compressive force
9	A	Annular cross sectional area
10	δ	Length change
11	$\acute{\varepsilon}$	Compressive strain
12	Ra	Mean surface roughness
13	Ω	Ohm
14	Hz	Hertz

Abbreviations

No.	Symbol	Definition
1	TOC	Total occlusal convergence
2	FEA	Finite Element Analysis
3	SEM	Scanning Electron Microscope
4	IPA	Isopropyl alcohol
5	ICP	Intercuspal contact position
6	EMG	Electromyography
7	Av/UC	Average uncemented
8	Av/PC	Average partially cemented
9	Av/FC	Average fully cemented

Chapter 1: Introduction

The composite structure of a cemented cast restoration consists of three parts, the casting, luting cement, and tooth structure. The weakest of these materials is the luting cement.

The cement provides the retention for the crown, whilst the mechanism of retention provided by cement is dependent on its type. Some provide only mechanical retention while others give mechanical and chemical adhesion. For zinc phosphate cement, the mechanism of retention is by mechanical interlocking between irregularities on the fitting surface of the casting and the tooth preparation; there is no chemical adhesion between either the cement and the preparation nor the casting and the cement.

The geometry of a preparation affects the loads on the cement. Two important factors are the total occlusal convergence (TOC) and the axial height of the preparation. Jorgensen (1955) concluded that there was an inverse relationship between the retention of the crown and the total convergence angle, whilst Kaufman *et al* (1961) showed an increase in the retention as preparation height increased.

Durability of the cement is important not only for retention but in order to minimise microleakage and recurrent caries. Stress and strain on the cement lute might affect its performance. Functional occlusal loading can generate complicated stresses within the luting cement film between the restoration and its abutment. Such stress might lead to cement fracture, which can promote cement dissolution and induce clinical problems (Yamashita *et al*, 1998). Little experimental work has examined the behaviour of cement beneath crowns. The work that does exist in the literature has frequently used tensile or torsion loading. Non-axial loads produce complex stresses and primarily test resistance form, whereas, the use of axial loading facilitates investigation of the attachment or bond provided by the cement.

It would be helpful if data were available which could characterise the behaviour of the bond between the crown and the tooth provided by luting cement. The purpose of this work was to develop an *in vitro* model to allow study of the condition of the adhesive joint between a cemented crown and its die. It was proposed that the establishment of such a method would improve the understanding of the bond provided for dental restoration.

Chapter 2: Literature review

2.1. Ideal requirements for a cement lute:

Rosenstiel *et al* (1998) divided the ideal requirements of luting cements as the following:

2.1.1. Biological:

2.1.1.1 Biocompatibility:

They stated that the ideal luting agent should have little interaction with body tissues and fluids, be non-toxic and have low allergic potential.

2.1.1.2 Caries or plaque inhibition:

An ideal luting agent would actively prevent caries at the restoration tooth inter-face. Additionally it should possess antimicrobial properties that combat bacteria on the prepared tooth and reduce the effect of further plaque colonies at the restoration margins. However, the benefits of these formulations have yet to be reported clinically.

2.1.1.3 Microleakage:

An ideal luting agent would resist microleakage of bacteria and their toxins around dental restorations. Zinc phosphate, polycarboxylate and glass ionomer cements were equally suited for permanent cementation of restorations. Resin luting agents showed high initial leakage, indicating that it was not as desirable for permanent cementation purposes. The zinc oxide-eugenol cement showed increased leakage with time but was well suited for its indicated purpose as a temporary cement (Mash *et al*, 1991). Adhesive resin systems however have reduced micro-leakage in *in vitro* testing (White *et al*, 1995) and also under *in vivo* testing (White *et al*, 1994).

2.1.2. Mechanical:

It was emphasised that the ideal luting agent should have sufficient mechanical properties to resist functional forces over the lifetime of the restoration. Rosenstiel *et al* (1998) added that an ideal luting agent should resist degradation in the oral environment and adhere to underlying dentine over the life time of the restoration.

They subdivided the mechanical properties into the following:

2.1.2.1. Strength:

The luting cement should have sufficient strength to resist fracture and also long- term cyclic fatigue stresses.

2.1.2.2. Low solubility:

An ideal luting agent should be impervious to oral fluids and resist dissolution.

2.1.2.3. Water sorption:

An ideal luting agent should have low water sorption, since this property may influence the dimensional changes of the set cement.

2.1.2.4. Adhesive:

A reliable adhesive luting agent would enhance fixed prosthodontic treatment, improving retention through mechanical interlocking and chemical adhesion.

2.1.2.5. Low setting stresses:

Setting stress can lead to contraction gaps between the cement and the tooth and/or restoration surface. If these gaps were near the restoration-tooth interface microleakage could result. If the stresses were greater than the adhesive or cohesive strength of the luting agent, adhesion might be disrupted leading to failures (Verzijden *et al*, 1992).

2.1.2.6. Wear resistance:

Rosenstiel *et al* (1998) considered this property to be of lesser significant in traditional fixed prosthodontics. However, it is of concern with ceramic and composite inlays.

2.1.3. Aesthetic:

Rosenstiel *et al* (1998) reported that aesthetic properties of luting agents are of considerable significance with the increased use of translucent ceramic restorations, especially for the anterior restorations and with the increased demand for aesthetic dentistry.

The authors subdivided the aesthetic requirements into the following:

2.1.3.1. Colour stability:

Cement colour changes over time. This fact should be considered during design of the restorations. For instance, light cured resin is more colour stable than the dual-cured because of the amine accelerator in the latter, which is necessary for polymerisation (Brauer *et al*, 1979).

2.1.3.2. Radiopacity:

An ideal luting agent should be radiopaque to enable practitioners to distinguish between the cement line and recurrent caries as well as detect cement overhangs. Rosenstiel *et al* (1998) also stated that the luting agent should have radiopacity that is greater than dentine, so differentiation between secondary caries and marginal gaps on radiographs might become easier.

2.1.4. Working:

Rosenstiel *et al* (1998) subdivided the working properties into the following:

2.1.4.1. Film thickness, viscosity:

This property depends on the type of the luting agent. Generally, with increase in the cement thickness, the bond strength of the luting agent would decrease and both seating and retention of the casting might be questionable. This property is controlled by many factors, such as type of luting cement, mixing technique, amount of cement spacing, luting agent viscosity and manipulative variables.

2.1.4.2. Working time and setting time:

An ideal luting agent should have an adequate working time for mixing and manipulation. The setting time should be adequate to allow placement but not over long.

2.1.5. Manipulation:

The ideal luting agent should not be technique-sensitive and easily manipulated.

2.1.5.1. Temporary cement removal:

Ideally temporary cement should be removed easily. Temporary cements that contain eugenol have been shown to inhibit resin based luting cements polymerisation (Rosenstiel and Gegauff, 1988). Luting agents that contain eugenol should be avoided for cementing provisional restorations before resin bonding. Cleansing with pumice will leave a zinc oxide eugenol residue mixed with pumice, which can inhibit bonding. Etching with 37% phosphoric acid after cleaning with pumice may be the best way to remove ZOE (Rosenstiel *et al*, 2001). Recently, non-eugenol zinc oxide cements for temporary cementation have become available. These cements can also be used for patients who are sensitive to eugenol (Craig, 1997).

2.1.6. Cost:

Economics are of relevance; Rosenstiel *et al* (1998) reported that the traditional cements are considerably less expensive than the newer formulations.

2.2. Retention and resistance of crowns and bridges:

Masticatory and/or para-functional forces may dislodge cemented crowns. The stability of a restoration on a preparation has been ascribed to its retention and resistance features. Retention has been defined as the potential to oppose the removal of the restoration along its path of placement. Resistance however, has been defined as the capacity to prevent dislodgment of a crown by forces directed in lateral or oblique directions and prevents any movement of the restoration under occlusal forces. Retention and resistance are interrelated and often inseparable qualities (Shillingburg *et al,* 1997).

Kaufman *et al* (1961) defined the factors that influence casting retention; these are the tooth preparation, the casting, and the cementing medium.

1) The prepared tooth. This factor was subdivided into:

- 1.1. The surface area.
- 1.2. The height of the prepared surface.
- 1.3. The degree of convergence of the opposing walls of the preparation.
- 1.4. The texture of the surface of the preparation.
- 1.5. The intra-coronal retentive devices in a preparation.
- 1.6. The degree of retention provided by various components of the prepared area.

2) The casting. This factor was subdivided into:

- 2.1. The relative adaptation of the casting to the prepared tooth surface.
- 2.2. The texture of the fitting surface of the casting.

2.3. The effect on retention of individual castings in a series splinted and cemented simultaneously.

2.4. The properties of the cast metal required to maintain the cement seal.

3) The cementing medium. This factor was subdivided into:

3.1. The type of cement.

3.2. The effect of a planned opening in the casting.

3.3. The viscosity of the cement.

3.4. The seating force.

3.5. The duration of the seating force.

3.6. The time between the cementation and any unseating forces.

3.7. The angle of any unseating force.

3.8. The magnitude of the lever arm of any unseating force.

3.9. The compressive and shear strengths of the cementing medium.

2.2.1. Retention and taper:

Taper is an important parameter in crown preparations. The angle of convergence of the prepared tooth has a great influence on the retention of the crowns. Taper describes the gradual decrease in width of an elongated object. It is the relationship between two opposite and gradually converging (external) or diverging (internal) vertical surfaces of the tooth. It is the main feature of the geometry of a tooth preparation. Confusion resulting from the misuse of the term by describing the relationship of one wall in relation to the long axis of the preparation has led some authors to describe the relationship of two walls as the total angle of convergence (TOC).

Theoretically, the more nearly parallel the opposing walls of a preparation, the greater the retention, with the most retentive preparation being one with parallel walls. However, parallel walls would make fabrication and cementation of a crown virtually impossible. Surprisingly, there is no close agreement between authors in their recommendations as to the ideal taper. Historically, the recommended TOC ranged between 4^{o}-14^{o} (Rosenstiel *et al*, 2001). The literature supports a minimal convergence angle. Both Jorgensen (1955) and Kaufman *et al* (1961) demonstrated that the retention decreased as the taper increased.

Jorgensen (1955) established the relationship between total occlusal convergence (taper) and the retention of cemented crowns. He used a cone of galalith representing a natural tooth of 8mm diameter and 8mm height; the vertical angle of the cone (the convergence angle) had values of 5^{o},10^{o},15^{o},20^{o},35^{o} and 45^{o}. For each vertical angle, 10 cones were made. Truncated brass caps were used representing the cast crowns and for each vertical angle 3 caps were made. The cones and the caps were provided with handles to help in the tensile testing. The cementation pressure was 4kg maintained for 10 minutes.

The retention between the oblique surfaces of the cone and the cap was measured 24 hours after cementation by pulling them apart. The results showed that as the angle of convergence increased, the load needed to separate the cap from the cone decreased, which was represented mathematically.

Kaufman *et al* (1961) confirmed Jorgensen's results. They used dies made from aluminum alloy, turned on a lathe to represent full crown preparation. They prepared dies with different axial heights of 4, 7 and 10mm with opposing wall convergence of 1°,5°,10°,15° and 20°. The preparations that were utilised had various diameters from 0.175 to 0.375 inch.

The results of the study showed an increase in the retention as height increased. However the increase was not uniformly proportional.

- Each unit area of a prepared tooth surface had a comparable retentive ability, regardless of the other dimensions of the preparation.
- There was a linear increase in retention as the preparation increased in diameter.
- In more convergent preparations, the areas closer to the gingival margin contributed a greater proportion of the retention.

Many teachers have advocated a convergence angle of 5° to 7° when preparing teeth to receive crowns (Shillingburg *et al,* 1997), whilst Tylman and Malone (1978) reported that "the axial reduction should not exceed 10° to 14°". In clinical situations, however operators do not achieve ideal taper. Many studies demonstrated that clinicians tend to prepare abutments with total occlusal convergence angles (TOC) which are greater than those recommended. Dies taken at random from commercial laboratories by Eames *et al* (1978) were found to have an average taper of 20°.

Kent *et al* (1988) evaluated the degree of taper of 418 dies of preparations made over a period of 12 years by an experienced operator. They found that the mean taper of the preparations ranged from 8.6° to 26.6° depending upon the location in the mouth and visual accessibility. They found a mean of 15.8° between mesial and distal walls and 13.4° between facial and lingual walls for preparations in all areas of the mouth with a mean of 14.3°. The lowest taper was observed on 145 anterior teeth with a mean of 9.5°, whilst the greatest convergence was found for 88 mandibular full crowns with mean of 22.2°.

It has been determined that the TOC produced by dental practitioners, dental students, general practice residents, and prosthodontists is not usually ideal. Instead, the angles range between 12° to 17°; Smith *et al* (1999), Noonan and Goldfogel (1991), Nordlander *et al* (1988). Ohm and Silness (1978) measured microscopically the bucco-lingual and mesio-distal convergence angles of vital and root filled teeth on 190 stone dies prepared for complete crowns with acrylic resin facings. Teeth were prepared by students in their last clinical year. They found that the mean total convergence in vital teeth was approximately 19° to 27°, whilst the root filled teeth were between 12° - 37°. Weed *et al* (1984) showed that senior students prepared clinical complete crowns with a mean convergence of more than 24° and in typodont teeth, more than 13.51°. After the results were explained and suggestions given by the instructors, another test was given to the same students; the mean tapers improved to 22°. Mack (1980) concluded that the taper in practice was approximately 17° which is far from those advocated in textbooks.

Dodge *et al* (1985) affirmed that resistance was more sensitive to changes in TOC than retention when tested using cemented crowns on metal dies. They investigated the tipping resistance of crowns on teeth with TOC angles of 10°, 16°, and 22° that were 3.5mm in occluso-cervical dimension and 10mm gingival diameter. They showed that considerably less tensile force was required to remove the crowns along the long axes of the preparations than to dislodge them with forces other than along the long axes of the preparations. It was 30% at 22°, 22% at 16°, and approximately 19% at 10°. They also found that 22° TOC provided inadequate resistance as it was the only casting dislodged without deformation. They concluded that clinically 10° TOC was not easy to produce and suggested that 16° as the best convergence angle among those tested. Additionally, crowns can be easily removed by axial tensile rather than obliquely directed forces.

Goodacre (2004) reported that there are a number of factors that affect the amount of TOC and make it more difficult to achieve 10°-20°. Mandibular tooth preparation leads to greater TOC teeth compared with that of the maxillary teeth, Smith *et al* (1999), Nordlander *et al* (1988), and Kent *et al* (1988). Kent *et al* (1988) added that mandibular molars have been identified as being prepared with the greatest TOC. These teeth also showed the greatest TOC on mesial-distal surfaces. A study by Annerstedt *et al* (1996) however, reported that the greatest TOC were prepared facio-lingually compared with the mesio-distal surfaces. Mack (1980) on the other hand reported that the abutment teeth for fixed partial dentures were generally prepared with greater convergence than teeth prepared for single crowns.

Visual assessment of taper can also influence TOC. Mack (1980) emphasised that monocular vision produces greater convergence than the use of both eyes (binocular vision). Goodacre (2004) reported that during clinical tooth preparation, an occlusal view is frequently used to assess TOC. He reported however, this view is of limited value because small differences in the TOC are difficult to assess viewing occlusally. Instead, facial or lingual views are more valuable and can be assessed by evaluating the image of the preparation in a mouth mirror. Further, Goodacre (2004) developed a total occlusal convergence gauge sheet at Loma Linda University for students and practitioners to aid evaluation and measurement of convergence angles. Placing a die of the prepared tooth so the axial walls can be superimposed over the lines present on the drawing permits a close approximation of the TOC. Goodacre *et al* (2001) reported that after anatomic reduction, most teeth have specific geometric forms when viewed occlusally. Prepared mandibular molars are rectangular in shape; maxillary molars are rhomboidal, whilst premolars and anterior teeth frequently are oval. They added that these geometric shapes have traditionally provided resistance to dislodging forces on individual crowns and fixed partial dentures. Hegdahl and Silness (1977) compared the area that created resistance form on conical and pyramidal tooth preparations. They reported that pyramidal tooth preparations provided increased resistance because they possessed corners compared with the conical preparation. They concluded that the use of large convergence angles resulted in small resisting areas and should be avoided. Goodacre *et al* (2001) affirmed the importance of preservation of the facio-proximal and linguo-proximal corners of a tooth preparation. They added that teeth lacking natural circumferential morphology variation after tooth preparation (round teeth) should be modified with the creation of grooves or boxes in the axial surfaces.

The literature and clinical experience confirm that small convergence angles are difficult to produce and somewhat greater convergence angles seem to be satisfactory clinically.

Wiskott *et al* (1996) investigated the relationship between total occlusal convergence angle (TOC) and the resistance of cemented crowns subjected to dynamic loading. Total occlusal convergence angles of 2.5°, 5°, 10°, 15°, 20°, 30°, and 40° were used for the full crowns which were cemented with zinc oxide-eugenol, zinc-phosphate, glass-ionomer, or resin composite cements.

Dynamic stresses were applied to the luted crowns using the staircase technique until the bond failed or reached one million load cycles. The relationship between convergence and resistance was approximately linear for all the cements tested. Crowns luted with composite resin cement were more resistant to dynamic lateral loading than those placed using glass-ionomer or zinc-phosphate cements. Crowns luted with zinc oxide-eugenol cement presented the least resistance to cyclic lateral stresses.

El-Ebrashi *et al* (1969) investigated in stress analysis studies the cement interface between the preparation and the restoration. They emphasised that the optimum taper is between 2.5° to 6.5° to minimise stress, with only a slight increase in stress as the taper increased from 0° to 15°. At 20°, the stress concentration was found to increase sharply.

2.2.2. Retention and preparation's surface roughness and luting cement:

Although surface finish can be a critical variable in clinical performance, there is a dearth of information regarding surface characteristics of teeth prepared for artificial crowns. Ayad *et al* (1996) characterised teeth prepared for complete cast restorations using three representative types of rotary instruments. They found a statistically significant difference in the surface topography of prepared teeth. Mean surface roughness (Ra) was 8.6 and 6.8mm for teeth prepared with diamond and tungsten carbide burs respectively. Teeth completed with finishing burs appeared to result in a smoother surface.

In a further article Ayad *et al* (1997) studied the relationship between surface characteristics of extracted teeth prepared for complete cast crowns and the retention of their respective cemented restorations. Retention was evaluated by measuring the tensile load required to dislodge the artificial crowns from tooth preparations with an Instron testing machine. The greatest retentive value (372.9 N) was for tooth preparations refined with carbide burs and cemented with a resin cement. However, the least retention value (201.6 N) was for tooth preparations completed with finishing burs and luted with zinc phosphate cement.

Wiskott *et al* (1999) presented a study to assess the effect of cement film thickness and surface texture (roughness) on the resistance of cemented crowns to dynamic lateral loading. Crown and abutment analogues were cemented using zinc oxide-eugenol, zinc-phosphate, glass-ionomer, and composite cements. The 3 degrees of surface texture subjected to investigation were (1) polished with up to 4,000-grit paper, (2) sanded using a 1,000-grit paper, and (3) sandblasted with 50 micron aluminum oxide. Testing was conducted according to the staircase procedure. The specimens were subjected to rotational fatigue loading until the cement bond failed or the components reached 1,000,000 stress cycles.

For zinc oxide-eugenol, zinc-phosphate and glass-ionomer cements increasing surface texture had a moderate effect. For composite resin cement, sandblasting doubled the resistance to dynamic lateral loading. For both parameters tested (cement thickness and surface texture), the ascending order of resistance was zinc oxide-eugenol, zinc-phosphate, and glass-ionomer cements. Crowns cemented with composite resin presented the highest resistance to dynamic lateral loading.

They concluded that texturing the surface of the abutment and the restorations using sandblasting increased the resistance to dynamic lateral loading.

2.2.3. Retention and occluso-gingival height:

Occluso-gingival height is an important factor for retention and resistance of crown and bridgework. The height of the prepared tooth affects resistance more than retention but the longer the preparation the more retentive the restoration. That is because of the greater surface area whilst the increased height interfaces more with the arc of the casting pivoting on the margin on the opposite side of the restoration (Smyd, 1944).

A further study by Wiskott *et al* (1997) investigated the relationship between tooth preparation height and diameter and the resistance of cemented crowns to dynamic loading. The results of the study showed that:

1. The relationship between abutment height (or diameter) and resistance to dynamic lateral loading was approximately linear.
2. Crowns luted using a resin composite cement were more resistant than those using glass-ionomer cement, which in turn were more resistant than crowns cemented with zinc phosphate or zinc oxide-eugenol cements.

Traditional retention in dentistry depends on geometric form rather than adhesion. Cements such as zinc phosphate are non-adhesive and act by increasing friction between the tooth and casting's fitting surface. Rosenstiel (1957) defined a closed lower pair of kinematics elements as a relationship between two bodies; one restrained the movement (tooth preparation) of the other (restoration). Mathematically, only three types of lower pairs could exist. These are turning, twisting, and sliding pairs. In dentistry, the last one could be applied to that formed by two cylindrical surfaces constrained to slide along each other. The pair could be constrained if the two cylinders were closed or shaped to prevent movement at right angles to the axis of the cylinder.

The relation between the occluso-gingival to bucco-lingual dimension has an important role in resistance of cemented restorations. Excellent resistance form could only be obtained easily when the length and diameter of preparation were optimal. When these are less than ideal, increased attention should be given to additional preparation features. Weed and Baez (1984) presented and validated a diagram for determining various tapers of preparation that provided resistance form based on the occluso-incisal to inciso-cervical dimensions and cervical diameters of 10mm. They used 50 stainless dies of complete cast crowns which subdivided into five groups with 3.5mm axial height and 10^{o}, 13^{o}, 16^{o}, 19^{o}, and 22^{o} total occlusal convergence. Gold copings were cast for each die.

A displacement force was applied to the castings that simulated a 30^{o} cuspal incline until the castings were deformed marginally except the 22^{o} group which displaced without deformation. They concluded that at 22^{o}, which represented 0.35 ratio, preparations lacked resistance form which had been predicted by the diagram. They recommended that the occlso-cervical to facio-lingual ratio should be 0.4 or higher. However, a continuous maximum load was applied until the castings were dislodged. This was not considered to be as representative of a clinical failure because oral occlusal forces are not sufficient to deform complete crown castings to the extent found in this study.

Parker *et al* (1991) evaluated the resistance of 294 single artificial crowns to dislodgement from their dies. This was carried out by grasping the casting between thumb and index finger and attempting to roll the un-cemented casting off the die with finger pressure. This seems a rather empirical test. The force was applied buccaly, lingually, mesially, and distally from a fixed point of rotation. They found that 96% of incisors, 92% of canines, 81% of premolars, and 46% of molars showed adequate resistance despite the variations in the prepared tooth form and dimensions.

2.2.4. Retention and die relief:

Die relief is the most common method of achieving sufficient space between metal castings and tooth surfaces for cement to flow during seating and to give space accommodating the cement lute layer. The application of die spacer to crown preparation dies prior to the fabrication of the cast crown is an acceptable procedure to improve the fit of the cemented restoration.

Studies have shown that the retention of the restoration will be improved, unchanged, or reduced when an appropriate thickness of die-spacer is used.

Lee and Ibbetson (2000) assessed the influence of die relief upon the seating and retention of cast full gold crowns constructed for natural human teeth. In half the samples, the dies were relieved by approximately 40 microns. Crowns were made by the indirect technique and were cemented with zinc phosphate cement in a static/dynamic jig. Seating behaviour and vertical discrepancies were measured electronically. Tensile bond strengths were determined in a universal-testing machine.

There were significant differences in seating between paired relieved and unrelieved samples ($p < 0.05$) whilst there were no statistically significant differences in tensile bond between paired relieved and unrelieved crowns.

Carter and Wilson (1996) investigated the effect of the number of die relief layers on retention. They prepared ten extracted molar teeth to a standardized complete crown preparation. Five stone dies were constructed for each tooth and coated with zero, two, four, six, and eight layers of a paint-on die spacer. Crowns were fabricated on each die and the force required for removal from each tooth was measured prior to and after cementation with zinc phosphate cement. The force required to remove the crowns before cementation decreased with increasing layers of die-spacer. Following cementation, the mean crown elevation decreased from 547 micrometres (zero layers) to 38 micrometres (eight layers); while the mean removal force increased from 250N (zero layers) to 375N (eight layers).

Passon *et al* (1992) however, evaluated the effect of increasing die spacer thickness on the retention of cast full crowns. Cast copings were fabricated on dies coated with either 0, 4, 8, 12, or 16 coats of die spacer, cemented on the respective teeth and removed using an Instron Universal Testing Machine. There were no statistically significant differences (P </= 0.05) between the mean force required to remove the cemented copings.

2.2.5. Retention and adhesive strength of luting cements to casting alloys:

There are different types of alloys that can be used as retainers in fixed prosthodontics work. Each alloy has specific characteristics according to its mechanical and physical properties.

Previous studies have shown that glass-ionomer and adhesive resin cements can bond to various alloys, while zinc phosphate cements cannot. Saito *et al* (1976) investigated the adhesive strengths of four commercial polycarboxylate cements to five dental casting alloys and compared these with zinc phosphate cement. The following results were obtained:

(1) The polycarboxylate cements showed adhesion that was 4 to 12 times greater than that of the zinc phosphate cement to all alloys tested.

(2) The adhesive strength of the polycarboxylate cements was greater to the chemically active substitute alloys, such as the copper, nickel-chromium, and silver-tin-zinc alloys. The adhesion of the polycarboxylate cements to the chemically stable gold and silver-

palladium alloys was not as great, but was four to six times that of the zinc phosphate cement.

(3) The differences in adhesive strengths between the brands of polycarboxylate cement were generally slight or statistically insignificant.

Mojon *et al* (1992) evaluated the bonding properties of three luting cements during the first seven days after cementation. Thirty cylinders were cast with high noble porcelain fused to metal (PFM) alloy and luted in pairs with one of the cements before being fractured in shear testing. The results showed that zinc phosphate cement was the weakest material, whereas the adhesive resin was the strongest. The glass-ionomer cement reached its maximum bond strength after two days, whereas storage time had no influence on the zinc phosphate cement. The adhesive resin cement was slightly, but not significantly, weaker after one week in water.

2.2.6. Retention and type of luting cement:

Browning *et al* (2002) compared the retentive strengths of resin, glass-ionomer, and zinc phosphate cements under adverse conditions. They mounted and prepared thirty extracted teeth in their long axes. The axial wall height was 3mm and the convergence angle was 28° to increase stress on the cement. Copings were made and cemented either with resin cement, glass-ionomer cement or zinc phosphate cement. The copings were loaded in tension, and the amount of force needed to remove the coping was recorded. They reported that the mean load required to remove the copings was, 9.4 MPa for resin, 5.0 MPa for glass-ionomer, and 3.1 MPa for zinc phosphate cements. They concluded that the resin cement was significantly stronger than both the glass-ionomer cement and the zinc phosphate cement when used on these short tapered preparations.

2.2.7. Retention and cement film thickness:

Mayhew *et al* (1982) reported that cast gold crowns cemented with a film thickness of 44 μ were significantly more retentive than those cemented in the range of 113-225 μ.

Jorgensen and Esbensen (1968) worked with crown and abutment analogues cemented with zinc phosphate cement and subjected to tensile stress. They reported that the increase in the film thickness from 20-140 μ lowered the retentive strength of veneered crowns by approximately 33%.

In addition, adhesive technology has demonstrated that thin joints are considerably stronger than thicker ones. Alster *et al* (1995) investigated the effects of layer thickness and contraction stress on the tensile strength of resin composite joints. They cured cylindrical samples of a chemically initiated resin composite (Clearfil F2) in restrained conditions and subsequently loaded them in tension. Tensile strength decreased gradually from 62 +/- 2 MPa for a 50 μ layer to 31 +/- 4 MPa for a 2.7mm layer. They found the types of failure were exclusively cohesive in resin for layers between 50 and 400 microns thick. The failures were cohesive or mixed adhesive/cohesive for 500-700 microns, whereas the failure was always mixed adhesive/cohesive for 1.4-2.7mm layers. For the resin composite tested, the contraction stress did not endanger the cohesive strength. They concluded that if adhesion to tooth structure were improved, thinner adhesive joints might enhance the clinical success of luted restorations.

Chana *et al* (1997) also reported a decrease in mean tensile strength of a resin-nickel-chromium alloy composite structure with an increasing cement thickness for two resin cements. They evaluated the influence of varying the cement thickness upon the tensile strength of nickel-chromium specimen cylinders cemented with Panavia Ex or Panavia 21 resin cements.

The exception from the trend towards increasing film thickness leading to a decrease in the retention of cemented crowns is polycarboxylate cement. McIntyre *et al* (1994) examined the bond strength of polycarboxylate cement when used to cement gold alloy specimens to dentine. The film thickness varied over a range of 10-150 μ. Tensile and shear bond strengths were measured and showed an increase in bond strength as the film thickness increased.

2.2.8. Resistance and the cementing load:

Proussaefs (2004) evaluated the effect of different cements on resistance to dislodgment of crowns cemented on preparations lacking geometric resistance form. The author prepared Ivorine teeth using a milling machine with no geometric resistance form, 20° total occlusal convergence, 0.9mm wide shoulder finish line, and a 2.5mm axial wall height. Ten metal test specimen die replicas and 10 standardized metal crowns with recipient sites for the application of external forces through a universal testing machine were fabricated. The crowns were cemented on the dies under 5 and 10kg external loads, the marginal openings measured, loaded to dislodgment, and cleaned of cement. The process was repeated using zinc oxide-eugenol (ZOE), zinc phosphate (ZPh), resin modified glass ionomer (RMGI), and composite resin (CR) cements.

The oblique dislodgment forces, measured with a Universal testing machine, were 40.18N (+/-6.76) for ZOE, 215.65N (+/-45.79) for ZPh, 165.43N (+/-19.53) for RMGI, and 181.54N (+/-30.75) for CR, when crowns were cemented under 5kg loads. The corresponding values for 10kg loads were 38.62N (+/-4.19), 274.86N (+/-54.22), 139.70N (+/-21.71), and 160.40N (+/-21.21) respectively. He found that only zinc phosphate cement produced statistically enhanced resistance when crowns were cemented under 10 kg force ($p= 0.035$).

In summary, most investigations of the cement lute have examined retention rather than resistance form. In the real life however, the cemented cast restorations are tested frequently with regard to their resistance.

2.3. Cement film thickness under castings:

Phillips (1996) reported that during the cementation procedure, the luting cement is forced into the surface irregularities of the tooth and casting, so that the set material is retained purely mechanically and/or chemically with the latter dependent on the type of luting material. The strength of the retention is therefore dependent in part on the strength of the interlocking tags of the cement. A thin cement film is believed to have fewer internal flaws compared with a thicker one.

Dimashkieh *at al* (1974) described a conventional procedure to measure the film thickness of luting cements under castings. The crown is cemented into position on its die, after setting the assembly is sectioned vertically to allow measurement of the cement film thickness.

Reduced cement film thickness can improve crown seating and decrease marginal discrepancies. Improved marginal adaptation has the potential to reduce plaque accumulation, periodontal disease, and cement dissolution (Yu *et al*, 1995).

Craig (1997) reported that ANSI/ADA Specification No.96 has requirements for cements designed for the seating of precision appliances. The maximum film thickness is 25μm. He added that the film thickness of zinc phosphate cement greatly determines the adaptation of the casting to the tooth. In a comparative way, zinc oxide-eugenol cement has a greater film thickness than is desirable; however, the products on the market today are able to meet the requirement that the film thickness be not more than 25μm as determined by the specification test. The zinc polycarboxylate cement on the other hand, has a film thickness that is slightly higher than of the zinc phosphate cement but is well within the clinical and specification test limits (25-48 microns).

Oilo (1978) has recommended that the film thickness should be above the maximum grain size of the powder, ideally between 25μm and 50μm. However, McLean and Von Fraunhofer (1971) suggested that a film thickness of up to 120μm at the gingival margins of a casting can be successful clinically.

A comparative study to determine and compare the film thicknesses of different luting agents was carried out by White and Yu (1992). The method was in compliance with American National Standards Institution/American Dental Association (ADA) Specification No. 8 for zinc phosphate cement. Each of the 20 materials tested was manipulated exactly as described in the manufacturers' instructions. An electronic gauge with an accuracy of 0.5 micron was recalibrated after each recording, and each luting agent was measured 10 times. Nine materials satisfied the ADA type I specification for film thicknesses less than 25 microns, and these included hydroxyapatite, glass ionomer, zinc phosphate, and polycarboxylate cements. Five other materials met the ADA type II specification for film thicknesses of less than

40 microns, and these included some glass ionomer cements, resinous cement, some zinc phosphate cements, and glass ionomer-resinous hybrids. Six resinous cements recorded film thicknesses greater than 40 microns and suggestions were made regarding future development and research.

2.3.1. Factors affect the cement film thickness beneath cast restorations:

Jorgensen (1960, a) reported the effect of various factors on the film thickness of cement under cast restorations. Cementing force, duration of the force, viscosity of the cement mix, the taper of the preparation, and venting through the occlusal surface were compared.

The factors that can affect the film thickness of the luting cement under the casting are:

1. Viscosity of the mixed cement.
2. Pressure of the cementation procedure.
3. Taper of the preparation.
4. Venting of the casting and its internal relief.
5. Vibration during cementation.
6. Marginal configuration.

2.3.1.1. Viscosity of mixed cement:

Viscosity is a measure of consistency of a fluid and its inability to flow due to the internal frictional force. A viscous fluid flows slowly. The manipulation of particularly acid-base cements has a marked influence on their viscosity, and the variable factors are the powder-liquid ratio, the temperature, the rate of mixing, and the particle size distribution of the powder. Incorporating more powder during mixing, increasing ambient temperature and using a rapid spatulation technique will lead to a more viscous mix and ultimately to a thicker lute. Conversely, using the recommended powder-liquid ratio, cooling the glass slab just before use and incorporating small increments of powder will generally produce a thinner film, although the optimum manipulative features differ from one material to another.

Rheology is the science of flow or deformation of materials and can be applied to both solids and liquids. It may be regarded as the effect of internal frictional forces of a liquid resisting a force acting to make it move. Different materials exhibit different rheological behaviour; where viscosity is constant and independent of shear rate, the material is said to be Newtonian. If viscosity increases with increasing shear rate, the material is said to be dilatant, and if viscosity decreases with increasing shear rate, the material is said to be pseudoplastic (McCabe and Walls, 1998).

The rheology of dental cements is further complicated by a rapid increase in rigidity within few minutes after mixing as the material solidifies, therefore all cements will show an increase in viscosity with time.

The rheological behaviour of five luting cements was compared by Lorton *et al* (1980), and differences in rheological behaviour were illustrated. Using apparatus which was essentially a piston and cylinder; it was shown that zinc phosphate cement behaved in a Newtonian fashion whilst polycarboxylate was pseudoplastic. The clinical application of rheology suggests that when using a cement, which viscosity decreases with increasing shear rate, e.g. zinc phosphate, the use of rapid tapping forces would encourage extrusion of the cement from beneath the casting. If the cement viscosity increases with high shear rates, e.g. zinc polycarboxylate, it would be more effectively extruded by a slow, steady application of force.

2.3.1.2. Pressure exerted during cementation:

White *et al* (1992) examined the effect of seating force on the film thickness of different luting agents. The materials tested were zinc phosphate cement, glass ionomer cement, polycarboxylate cement, and resinous cement with a dentinal bonding agent. Each class of material was measured 10 times at six different seating forces. Analysis of variance and multiple comparisons testing disclosed that the seating force and class of material strongly influenced the film thicknesses of luting agents.

Clinically seating of crowns may be improved by additional mechanical aids, namely Merdart pressure applicator, orange wood block, and Burlew disk. Oliveira *et al* (1979) concluded that the Medart pressure applicator produced better adaptation. Orange wood and Burlew disks produced similar results, they were less efficient however. Cotton rolls resulted in the highest fit discrepancy. Wang *et al* (1992) reported that the orangewood stick and bite device had a similar effect on crown seating.

2.3.1.3. Effect of taper, venting, and internal relief:

These three factors are related because they affect the build up of hydrostatic pressure in the cement as the casting is seated.

Thinner cement film may be achieved by increasing the taper of the preparation and /or incorporating an occlusal vent to allow escape of excess cement (Kaufman *et al*, 1961) and (Dimashkieh *et al*, 1974). Placing a modified axial groove on the buccal aspect of the preparation after the impression has been taken will improve seating of the crown providing a cement escape way because the groove is not represented on the casting (Miller and Tjan, 1982). Axial grooves have the advantage of obviating the need for an occlusal vent in the crown, but they are less conservative of tooth tissue and are only useful in molar teeth which have more tooth structure.

Techniques of internal relief of cast restorations used by dentists since the early 1920s have included (1) internal carving of wax patterns, (2) internal grinding of castings, (3) aqua regia etching, and (4) electrochemical milling (Hollenbeck, 1928) and (Bassett, 1966).

A simpler method of providing cement space is by using die relief. Paint-on die spacer applied to the dies before waxing is presently advocated and used for internal relief. The film thickness of 20-40 μm, which is recommended in the literature, is built up by repeated coating of appropriate surfaces. A paint-on die spacing material is the most common method of achieving sufficient space between metal castings and tooth surfaces to allow cement to flow during seating. The material is used to coat the die to within 1mm of the margin. Various types of die relief can be used including model paint, coloured nail varnish, and polymers dissolved in volatile solvents.

2.3.1.4. Vibration during cementation:

Studies have indicated that dynamic seating methods can reduce seating discrepancies associated with zinc phosphate and resin cements. Yu *et al* (1995) investigated the effect of three loading methods on the film thicknesses of six luting cements compressed between glass plates. The cements were zinc phosphate, resin-modified glass ionomer, encapsulated glass ionomer, adhesive composite resin, composite resin, and polycarboxylate. The method was derived from American Dental Association specifications for cement film thickness. In control groups, the cements were placed between two glass plates and statically loaded with a 15kg weight. The test groups were initially similarly loaded, and then for 30 seconds further subjected to simulated repeated patient opening and closing, vibrations from an electromallet, or an ultrasonic device. They found that the mean film thicknesses ranged from 7.4 micrometres for polycarboxylate / ultrasound up to 40.3 micrometres for composite resin / static. Two-way analysis of variance showed that the effects of material type and cementation method and their interaction all significantly affected film thickness ($P < 0.0001$). Multiple range analysis showed that dynamic methods were generally superior to static loading and that the ultrasonic method was the best overall. They concluded that the different dynamic loading methods all significantly decreased cement film thicknesses between glass plates. The ultrasonic method was the most effective and composite resins were more affected than other materials.

To assist in the final seating of a casting, static loading or dynamic loading (vibration) may be used. An *in vitro* study (Koyano *et al*, 1978) compared six combinations of static and dynamic loads during cementation of inlays and crowns. They found that dynamic pressure produced thinner cement films than static pressure, and that vibratory pressure was more effective than the automatic mallet. Vertical vibration was found to be more effective than horizontal vibration, and the combination of static pressure followed by dynamic pressure produced thinner cement films than any individual technique.

Chan and Setchell (1997) however suggested that vibration reduced the vertical discrepancy between a full crown and a die but was only effective when an escape way of some kind was provided either by die relief or occlusal venting.

2.3.1.5. Marginal configuration:

Piemjai (2001) investigated the marginal discrepancy and retention with various margin designs. He reported that the shoulder and shoulder with bevel finish lines provided better crown retention than the chamfer, whilst glass-ionomer cement provided greater crown retention than zinc phosphate cement. However, further analysis revealed no significant correlation between marginal seating and crown retention.

Silness and Hegdahl (1970) studied mathematically the effect of the geometric configuration of the finishing line of the preparation and the crown margin on the surface area of exposed cement to find out which type of preparation gave the least exposure of cement. They found that the exposed cement may amount to several square millimetres and that the greatest area was found with 90° shoulder preparation and the least with chisel edge and bevelled shoulder preparations. This study did not relate these to possible differences in seating with various type of finish line.

However, Gavelis *et al* (1981) investigated the relationship between the marginal configuration and the occlusal seating of full crowns. They concluded that the best occlusal seating was found with 90° shoulder preparation, and that a 90° shoulder preparation with a parallel bevel gave the poorest occlusal seating. This was attributed to a build-up of hydrostatic pressure in the cement during seating. This begins earlier with parallel bevelled shoulder margins and therefore offers more resistance to seating than the 90° shoulder margins. With the latter, cement escape is possible until the crown is almost fully seated; the hydrostatic pressure is much less, thus permitting greater seating of the restoration.

2.4. Cement solubility:

Restorations, which are constructed outside the patient's mouth, are subsequently permanently cemented in the mouth via luting cement to provide retention and to fill the space between the restoration and the prepared tooth. One of the important characteristics of luting cements is that they should not dissolve, erode, nor should they leach important constituents into the oral cavity.

Clinical evaluation of these luting cements is not easy because much of the material is covered by the restoration. However, there will be inevitably an exposed cement line at the restoration margin. Breakdown of the cement matrix at the margin may lead to failure of the restoration as a result of the initiation of secondary caries in the marginal defect or with continued dissolution of the cement. Phillips (1996) reported that with the exception of resin cements, all have the potential for significant degradation in oral fluids. As a result; fragments can be lost from beneath a restoration, leakage and bacterial invasion ensue with subsequent problems of sensitivity, dental caries or both.

McCabe and Walls (1998) defined solubility as the measurement of the extent to which a material will dissolve in a given amount of fluid, such as water or saliva. They also added that the definition of solubility should not be confused with the definition of erosion, which combines the chemical process of dissolution with a mild mechanical action. The authors reported that American Dental Specification number 8 provides standard criteria for acceptance or rejection of dental cements according to their solubility in distilled water which does not replicate clinical conditions.

Mesu (1982) described a dynamic method to measure degradation of luting cements *in vitro*. He concluded that degradation is a complex process and appeared to follow a sequence of absorption, disintegration and dissolution. Factors, such as thickness of the cement layer, molarity, and pH of the medium affect the rate of degradation. Though the clinical predictability of this test method has not been determined, it gives the opportunity to measure the various stages of degradation and to define the different factors affecting this process.

2.4.1. Solubility tests:

It seems imperative that a dependable test be devised to predict the durability of dental cements. However no such method exists and solubility tests can be divided into *in vivo* and *in vitro*:

2.4.1.1. *In vivo* tests:

Norman *et al* (1969) compared the intra-oral solubility of three cements, which were placed in receptacles in removable partial dentures. These cements were subject to dissolution in oral fluids as well as to abrasion due to daily tooth brushing and tongue movement.

After measuring the loss of weight of these cement samples following 6 and 12 months in the mouth, they reported that silicate cement demonstrated the least loss with zinc phosphate cement intermediate and zinc oxide eugenol cement the greatest.

Richter and Ueno (1975) investigated the *in vivo* degradation of four dental cements (zinc silicophosphate, zinc phosphate, ZnO-EBA with alumina, and zinc polycarboxylate). The cements were placed in cavities prepared in the pontics of temporarily cemented fixed partial dentures. Two different locations were used to evaluate the influence of solubility alone and abrasion and solubility combined. The results showed that silicophosphate cement was most durable to wear or dissolute. In a comparative order the zinc oxide eugenol was more durable than ZnO-EBA cement and the zinc polycarboxylate cement. However, it was less durable than that of zinc silicophosphate cement. The degradation rate of both ZnO-EBA and zinc polycarboxylate cement was equal. The cements exposed to both wear and dissolution exhibited greater degradation than those subjected to dissolution alone, this difference was significant. However, the abrasion component in this study was high compared to the likely abrasion of cements retaining a cast restoration.

Pluim *et al* (1984) measured the dissolution rate of dental luting cements longitudinally. Cement samples in enamel were placed in a full prosthesis and worn in the mouth for up to 6 months. At given intervals, replicas of the cement surfaces were made and quantified by means of SEM stereographic photographs. With a stereometer the distance between cement surface and original enamel surface could be determined quantitatively. The results showed *in vivo* solubility rates for zinc phosphate and glass ionomer cement of about 80 microns and 2 micrometres per week respectively.

There is doubt in the literature about the validity of *in vivo* solubility experiments on dental cements. Mitchem and Gronas (1978) evaluated a method of comparing the *in vivo* solubility of permanent luting cements in 15 patients. The cements were carried in crypts placed in complete dentures; each holder contained six holes of 2mm in diameter and 2mm deep. A vented cover allowed for free access of fluids. The patients were instructed to clean their dentures normally but not to force bristles or other objects through the holes and to keep their dentures in water at night with no commercial cleansing agents being used. Six cements were evaluated and cement loss was measured at 6 months. The sample holders were removed, cleaned and an impression was made. The amount of cement loss was determined by measuring the amount of impression material protruding from the base under a Gaertner micrometer slide microscope.

The glass ionomer (at two different powder-liquid ratios) and silicophosphate cements demonstrated an approximately equal solubility, which was less than for the other cements tested. The remaining cements demonstrated greater loss and were equally affected by the environment. The study appeared to present means of evaluating clinical solubility under oral conditions that were close to clinical usage, there being little or no abrasion. However, the method did not replicate the thin film of luting cement beneath crowns.

Wilson (1976) discussed the specification test for the solubility and disintegration of dental cements. He reported that, although it was useful for quality control, it gave an imperfect account of clinical durability. He added that it could not be used to compare different classes of dental cements and it had little meaning if the products of hydrolysis were insoluble or volatile.

2.4.1.2. *In vitro* tests:

Due to the difficulty of clinical test procedures for dental materials, the use of the *in vitro* test is the way to provide the economical method for comparative evaluation of those materials.

Design of an *in vitro* test that reliably predicts the relative potential for disintegration of cements in the oral cavity has met with considerable frustration. *In vitro* data rarely correlate with clinical experience or solubility data obtained from the mouth. For several years much effort has been made to develop a meaningful *in vitro* test and a number of parameters that influence test results have been defined. For simplicity, these tests can be divided into three methods:

2.4.1.2.1. Gravimetric method:

Yoshida *et al* (1998) compared the solubility of three resin cements currently in clinical use with three brands of conventional luting agents. The three resin luting cements, All-Bond C&B (AB, Bisco) Panavia 21 (P21, Kuraray), and

Super-Bond C&B (SB, Sun-Medical), and the three conventional luting agents, Elite Cement 100 (EC, zinc phosphate cement, GC), HY-Bond Carbo-plus Cement (HCP, polycarboxylate cement, Shofu), and Fuji I (FI, glass ionomer cement, GC). Two types of media, distilled water and pH 4.0 lactic acid solutions were prepared, in which specimens were stored for 30 days. They found that the four luting cements, EC, FI, AB, and P21 were more soluble in lactic acid solution than in distilled water. The resin luting cements were markedly less soluble than conventional luting agents when placed in the lactic acid solution. They found that the solubility increased linearly with immersion period.

However, Nomoto and McCabe (2001) reported that the gravimetric method was not suitable for water-soluble materials, because it is difficult to determine the effects of inherent water within the materials, but it is specified in the ISO standard for resin-based materials which contain no water.

2.4.1.2.2. Difference in weight loss:

This method is based on measurement of the residual weight of a solution in which the cement has been immersed, after evaporating the water. It is commonly used in laboratory tests of solubility and disintegration and is the basis of the ADA Specification number 8.

2.4.1.2.3. Measurements of depth loss:

This method is a measurement of the depth loss of the cement in a cavity filled with the cement, which is based on the impinging jet erosion test. The erosion jet test method employs a jet of aqueous acid impinging on the surface of the cement causing loss by combined chemical/mechanical action. This method can be used for comparison of different cements' solubility (Mesu, 1982& ISO 9917, 1991).

Beech and Bandyopadhyay (1983) proposed a method for accelerating the dissolution and erosion of 'acidic' dental cements which gave relative solubility in accordance with *in vivo* observations, in contrast to the conventional test. It involved subjecting 24 hours old specimens to a jet of dilute acid (lactic or citric) and determining weight loss. The relative ranking for 'acidic' cements in the apparatus and (*in vivo*) was glass ionomer less than silicophosphate less than zinc phosphate less than polycarboxylate. The method was not valid for reinforced zinc oxide eugenol (EBA) cements. It considered that the method was the basis of a new solubility or erosion test in national and international standards for 'acidic' dental cements.

Nomoto and McCabe (2001) evaluated an alternative erosion test for dental cements. The method was evaluated by comparing the extent of the erosion of three different types of dental cement which were zinc phosphate, zinc polycarboxylate and glass ionomer. The cements were evaluated by measuring the depth loss of the cement placed in a cylindrical cavity after immersion in lactic acid solution (pH 2.74) or lactic acid/sodium lactate buffer solution (pH 2.74).

Changes in the pH of the eroding solutions were monitored. Identical specimens were immersed in distilled water and the degree of hygroscopic expansion was also confirmed. Surface profiles were recorded using a profilometer at 1, 3 and 7 days and the height at the centre of the specimen was measured using the shoulders of the specimen-holder as fixed reference planes.

They found that depth loss and pH increased with time. The depth losses of all cements were considerably more in the buffer solution than in the acid solution. Differences in the eroded depth for the three different kinds of cements clearly emerged. The depth loss for polycarboxylate was more than that for zinc phosphate, which was more than for glass ionomer cement. Although hygroscopic expansion was observed for glass ionomer cement, the value after 24 hours immersion was negligible compared with the eroded depth. The results from this study appeared to correlate with those obtained using a jet erosion test and with published data on clinical performance and suggested that this simpler test was suitable for standardization purposes.

2.4.2. Comparison of test methods:

Swartz *et al* (1989) evaluated three *in vitro* tests on the basis of their ability to predict the relative resistance of various types of luting cements to disintegration in the oral environment.

The tests were:

(1) Weight loss.

(2) Reduction in size of cement films held between glass optical flats.

(3) Reduction in transverse strength.

Storage media evaluated were H_2O as a control, 0.1 acetic acid pH 2.7, 0.01 acetic acid pH 3.4 and 0.001 M acetic acid pH 3.8. The cements tested were zinc phosphate, zinc silicophosphate, zinc polycarboxylate, and glass ionomer.

The results showed that the changes that occurred with storage in water ranged from none to slight with no separation of materials other than polycarboxylate cement. Weight loss and transverse strength of polycarboxylate specimens decreased and they eventually exhibited swelling. Zinc phosphate cement lost weight more rapidly than the other cements. Complete disintegration of the specimens occurred after 40 weeks. The specimens absorbed fluids and there was obvious swelling.

These data were compared with previously obtained measurements on the *in vivo* disintegration rates of the same cements. The best correlation with *in vivo* disintegration was obtained by test No. 2, the reduction in size of the cement films held between glass optical flats when the specimens were subjected to 0.01 M acetic acid. They also concluded that the surface area of the cement exposed to the medium appeared to be a critical factor in the disintegration rate of certain cements.

2.4.3. Correlation between *in vivo* and *in vitro* tests:

The solubility or degradation of dental cements has been determined *in vivo* as well as *in vitro* in various ways. Mesu and Reedijk (1983) used the same test specimen *in vivo* and *in vitro* and determined the correlation between both situations. They used four commercial luting cements, zinc phosphate, polycarboxylate, zinc oxide eugenol and glass ionomer cement. They created a 20µm cement film between glass plates (7mm) which left the edges of the cement layer exposed. The specimens 15 minutes after mixing were placed in perforated Teflon trays which were slowly raised and lowered through the dissolution medium of 500 ml of lactic acid (0.02M) buffered with sodium lactate to pH 4 also containing *S. mutans*.

The *in vivo* experiment was conducted on 12 patients. They used two specimens of the same type in the location of the second molar on a lower denture. The patients were allowed to clean their dentures with brush and water but the use of cleaning agents were prohibited. The pH and number of bacteria in the saliva, on the denture and on the specimen were determined. Registration of the degradation was recorded photographically *in vitro* every 24 hours for seven days and *in vivo* once a month for six months.

The results showed that glass ionomer cement was the least degradable cement in both situations (*in vivo* and *in vitro*) while the most degradable cement was zinc oxide eugenol. There was a correlation between the *in vivo* and the *in vitro* test with lactic acid ($p<0.025$). No correlation was found between the results in the streptococcal medium and *in vivo*. There was also a positive correlation between degradation and the total number of bacteria (*S.mutans* and *Lactobacilli)* on the specimen.

They concluded that the number of bacteria as well as the number of acid producing bacteria such as *S. mutans* and *Lactobacilli* in contact with the exposed area of the cement affected the rate of degradation.

2.4.4. Solubility factors:

There are many factors that can affect the solubility of luting cements.

2.4.4.1. Powder to liquid ratio:

Osborne and Wolff (1991) studied the effect of the powder/liquid ratios of polycarboxylate cement on its solubility. The data suggest that higher powder/liquid ratios were preferable when minimal solubility is a primary requirement.

2.4.4.2. Mixing temperature:

Mitchem and Gronas (1981) in their *in vivo* study placed cement sample holders in crypts built into dentures. After 6 months the amount of cement loss was determined and it was concluded that mixing of zinc phosphate cement on cooled glass slabs in which the condensed moisture is incorporated in the mix did not appear to affect clinical solubility.

2.4.4.3. Maturation time:

Williams *et al* (1992) investigated the clinical durability of restorations made using water-based cements using the jet-test method. Glass-ionomer, zinc polycarboxylate, silicate and zinc phosphate cements were tested at 24 hours and 2 months. The glass ionomer cements showed a significant reduction in erosion rate with time and the zinc polycarboxylates showed some reduction but not to a significant degree. The older silicate and zinc phosphate cements samples showed increased erosion rates. They concluded that materials using polymeric acids had erosion rates which reduced significantly with time.

2.4.4.4. Protection from water and/or saliva contact:

Rodrigues Garcia *et al* (1995) evaluated the influence of protective agents (varnish and glaze-resin resin) on water solubility of two glass ionomer restorative cements (Chelon-Fil and Vidrion-R), using the gravimetrical test (A.D.A Specification No. 9).

They found that the two glass ionomer restorative cements tested required surface protection to avoid early solubility. Among the protective materials evaluated, the varnish was the most effective agent.

2.4.4.5. Time of water exposure:

Oilo (1984) investigated the effect of the time of exposure and the disintegration in water of various types and brands of cements (glass ionomer cements, polycarboxylate and zinc phosphate cement). He used gravimetrical measurement after exposure of the cements to a constant water jet. For zinc phosphate and polycarboxylate cements, no weight loss was observed in the period from 4 to 8 minutes after commencement of mixing. All the glass ionomer cements showed a significant loss of weight at 4 minutes and a somewhat reduced weight loss at 6 minutes after the start of mixing. Two cements, a filling and a luting material, showed reduced weight when exposed to a water jet even 8 minutes after the start of mixing. The early erosion as recorded in the present study conforms to the characteristic of the glass ionomer cements.

2.4.4.6. pH of the solution:

Iwaku *et al* (1980) compared three types of luting cements for acidity, disintegration and film thickness. The pH during setting of the two types of polycarboxylate cement exceeded 6 after 5 hours; that of the water-settable polycarboxylate cement was the highest, almost reaching neutral. The zinc phosphate cement reached pH 3.4 after 5 hours and 5.5 after 24 hours. All cements tested showed remarkably greater disintegration in the lactic acid solution than in the distilled water. The disintegration of the water-mixed zinc polycarboxylate cement in distilled water was about half that of the other cements. The disintegration in the lactic acid solution was approximately the same for all cements tested whilst the film thickness was lowest with the water-mixed zinc polycarboxylate cement.

2.4.4.7. Bacterial effect:

Mesu and Reedijk (1983) concluded that the number of bacteria as well as the number of acid producing bacteria such as *S.mutans* and *Lactobacilli* in contact with the exposed area of the cement affected the rate of degradation.

2.4.4.8. Exposure to abrasion:

Phillips *et al* (1987) reported that the rate of disintegration of the four cements occlusally compared to the cervical areas showed no significant difference. Richter and Ueno (1975) reported that the cements exposed to both wear and dissolution exhibited greater degradation than those subjected to dissolution alone.

2.4.4.9. Marginal width:

Guzman *et al* (1997) investigated the effect of marginal gap width on the wear resistance of the luting cement in areas where no occlusal contact was present. Three types of resin luting cement and one resin-modified glass-ionomer cement were used with two inlay systems, a resin composite and an all-ceramic system. Bovine enamel represented tooth structure. Toothbrush abrasion was the wear modality. Three predetermined gap widths were selected: 240 +/- 30 micrometres, 150 +/- 30 micrometres, and 60 +/- 30 micrometres. All specimens were thermo-cycled. Regardless of the luting cement or the restorative material, there was a significant difference ($p < 0.05$) in wear resistance of the cement among the three gap distances at both the enamel and restoration interfaces. Vertical wear of the luting cement at the enamel interface increased linearly with marginal gap distance when all four cements were considered together ($r^2 > 0.51$) regardless of type of restorative material used. The resin-modified glass-ionomer cement showed the least amount of wear for all variables considered. Significant differences in wear were found between the four luting cements at wide gap distances (240 micrometers) at the enamel interface, regardless of type of restorative material used. No significant differences were found between the two restorative materials at the enamel interface at the three gap distances.

Jacobs and Windeler (1991) investigated the rate of type I zinc phosphate cement solubility as it related to the degree of marginal opening. Standardized test samples were constructed that would simulate clinically relevant marginal gaps of 25, 50, 75, and 150 micrometres and their subsequent cement lines. The study was divided into two phases. Phase 1 evaluated the effects of simple diffusion on cement solubility in a static environment, whereas phase 2 investigated the effects of convective forces on cement dissolution in a dynamic environment. Both the phase 1 and phase 2 studies demonstrated no significant difference in the rate of cement dissolution for the 25, 50, and 75 micrometres test groups. The 150 micrometres test groups for both studies, however, demonstrated an increase in the rate of cement dissolution. The results of the phase 1 and phase 2 studies could not be compared because different methodologies were used.

Conclusions:

Clinical success of luting cements depends on parameters such as mechanical properties, biological effect on dental tissues and the solubility in the oral environment.

Cement solubility is an important aspect of luting cements. It is likely to have an effect on the survival rate of the dental restoration as a result of the seepage of the oral fluids and bacteria with their products between the restoration and the tooth tissue. This may lead potentially to recurrent dental caries, loss of pulpal vitality and ultimately failure of the restoration.

Resin cements have the least potential for significant degradation in oral fluids. Within the water-based luting cements, glass ionomer cement exhibited less solubility than zinc phosphate and zinc polycarboxylate cements, which are more or less equal in their solubility.

2.5. Bite force values and measurements:

Mastication consists of crushing or vertical forces and milling or translatory forces. The forces developed between the teeth are those produced in the normal chewing of foods and those that can be applied when a maximum biting load is applied without food being present.

The load/energy generated during mastication can be absorbed by the food bolus, teeth, periodontal ligaments and supporting bone. The modulus of resilience of dentine is greater than that of enamel, thus it is better able to absorb impact energy. Enamel is a brittle substance with a high modulus of elasticity, a low proportional limit in tension and low modulus of resilience. Enamel is supported by dentine with a significant ability to deform elastically; hence teeth rarely fracture under normal occlusion (Phillips, 1996).

The actual biting stresses during mastication are difficult to measure because of the dynamic nature of the movement. Bates *et al* (1975) reported that the measurement of forces in the oral cavity has evolved from the placement of measuring devices between the teeth to the use of sophisticated strain gauges. Later, radiotelemetry methods were used to avoid the need for wires issuing from the mouth. Waltimo and Kononen (1993) used a quartz force transducer that served as a sensory unit. A micro-processor produced a numeric result, shown on a liquid crystal display (LCD).

Shinogaya *et al* (2000) compared a technique (Dental Prescale System) using pressure-sensitive foils for recording of maximal jaw closing force with a conventional clinical measurement using a miniature force transducer. For assessment of load distribution the "pressure" image was superimposed on digitised images of dental casts and compared with clinical registrations of tooth contacts. The bite force recorded with foil was systematically higher than that recorded by conventional measurements. Further more teeth were assessed as being loaded than those indicated as having clinical occlusal contacts. The new technique was reported as promising although time consuming.

Phillips (1996) reported a number of studies that had determined biting forces. The Guinness Book of Records (1994) listed the highest biting force as 4337 N (975 pounds) that was sustained for 2 seconds. The average maximum sustainable biting force was approximately 756 N (170 pounds). Gateau *et al* (2001) studied the *in vitro* fatigue resistance of new types of glass ionomer cement used in a post and core investigation. They used 1.5 million cycles with a 400N masticatory load. They claimed that was equivalent of 4 years of normal mastication.

Craig (1997) reported that the maximum bite forces measured by strain gauges and a telemetric device ranged between 200-3500N. He also added that the repetitive stress occurring in restoration during mastication was about 300,000 times per/year. Braun *et al* (1995) studied bite force and its relationship with various characteristics in 142 dental students. They found that the mean maximum bite force was 738 N with a standard deviation of 209 N. In a further study, Gundler *et al* (1993) reported that maximum bite force was 700N and was applied 2 times / second (2Hz). Yamashita *et al* (1997) measured the strain on four-units bridges during function under an occlusal force of 200N.

Waltimo and Kononen (1993) used a quartz force transducer as a sensory unit. The maximal bite forces of healthy undergraduate dental students exceeded the values previously reported for unilateral housings. The mean maximum bite force value in the molar region was 847 N for men and 597 N for women.

2.5.1. Factors affect masticatory load:

Phillips (1996) reported there are different factors that can affect masticatory load; it varies markedly from one area to another in the same mouth and from one individual to another.

2.5.1.1. Area in the mouth:

Craig (1997) reported that the biting force depends on the location within the oral cavity; forces on teeth generally decrease from the molar to the incisor teeth. The biting force in the molar area was 400-800N, in the bicuspid and cuspid area 200-300N and in the incisor area was 150N.

Phillips (1996) also reported the same and found the masticatory load in the molar region to be between 400-890N (90-200 pounds), in the premolar region between 222-445N (50-100 pounds), at the cuspids between 133-334N (30-75 pounds) and at the incisors, 89-111N (20-55 pounds).

2.5.1.2. Ethnicity:

Shinogaya *et al* (2001) investigated the effect of ethnicity on the degree of clenching force and distribution of the load over the dentition. The maximal clenching force was measured in 12 young Danish females, 12 young Japanese females, 12 young Japanese males and 10 senior Japanese males using the Dental Prescale System. The arch width and average pressure in young Danish females were significantly smaller/lower than in Japanese females.

2.5.1.3. Gender:

There are many studies emphasising that bite force is gender-related. Braun *et al* (1995) reported that in relation to gender the mean maximum bite force was found to be statistically significantly different. Phillips (1996) reported the biting force is higher in males compared to females whilst Craig (1997) reported that the bite force in women was generally 90N less than for men. Waltimo and Kononen (1993) in their examination of undergraduate dental students found that the mean maximal bite force value in the molar region was 847 N for men and 597 N for women.

2.5.1.4. Age:

Phillips (1996) emphasised that the masticatory bite force was greater in young adults than children. Braun *et al* (1996) measured bilateral bite force in a sample of 457 subjects (231 males and 226 females) from 6 years through 20 years of age. The mean maximum bite force was found to increase from 78 Newton at 6 to 8 years to 176 Newton at 18 to 20 years. Shinogaya *et al* (2001) reported that the average pressure in young Japanese males was also greater than in senior males.

2.5.1.5. Nature of opposing tooth material:

Craig (1997) reported that biting force depends on the nature of the teeth. The occlusal force in patients who had a fixed bridge replacing a first molar was 250N on the restored side compared with 300N on the opposite side where there was a natural dentition. In patients with removable partial dentures however, the force measurements ranged from 67N to 235N. Additionally, patients with complete dentures have bite force in a range of 100N in the molars and bicuspids but 40N on the incisors.

2.5.1.6. Relation of mandible to maxilla:

The relation of the mandible to the maxilla affects the maximum biting force. Erhardson *et al* (1993) recorded the activity of the jaw elevator muscles (EMG) during maximal clenching in ICP (intercuspal contact position) , and unilaterally on 2nd and 1st molars, premolars, canines and central incisors in 10 subjects with complete dentitions and normal occlusal relationships without signs and symptoms of temporomandibular disorders. The EMG activities were converted to relative muscle force. The distribution of the relative muscle force on the teeth and condyles was then calculated by means of a three-dimensional mathematical model. They found that the magnitude of the muscle force was strongly associated with the localization of the biting point. The greatest muscle force was produced during clenching in the intercuspal position (100%). On unilateral clenching 75% of the ICP force was reached during rubber pad clenching on the 2nd molars, 85% on the 1st molars and 45% on incisors.

2.6. Stress and cement lute:

Smyd (1961) reported that the kinematics of dental restorations requires that all materials which go into their construction, namely, alloys, porcelain, plastic, dental cement, and tooth structure, be used only at loads which these materials can sustain elastically. The most critical element among this was the dental cement.

Functional occlusal loading can generate complicated stress within the luting cement film between the restoration and its abutment. Such stress may lead to cement fracture, which can promote cement dissolution and induce clinical problems (Yamashita *et al*, 1998).

Van Meerbeek *et al* (2003) stated that during their life-time, restorations are subjected to cyclic loading, each load is insufficient to provoke failure, but in the long-term can lead to restoration loss. They added that clinical trials are the ultimate test for dental restorations, but that the true reason for failure can not be differentiated due to the simultaneous impact of diverse stresses on restorations within the oral cavity. Laboratory testing can evaluate the effect of a single variable, while keeping all others constant. In general, laboratory testing is easy, fast and relatively cheap to screen new materials and/or techniques. Ideally, the final objective should always be predicting long-term clinical behaviour, though direct translation to the clinical situation is often difficult or even impossible.

Ways to determine the stress values and their distribution have been reported by Yettram *et al* (1976); these included:

1. Transmission and reflection two-dimensional photoelasticity (Mahler and Peyton, 1955) and (Craig *et al*, 1967).
2. Three dimensional photoelasticity (Johnson *et al*, 1968).
3. Brittle lacquers.
4. Electrical resistance of strain gauge technique (Tillitson *et al*, 1970).

Recently, Yamashita *et al* (1998) reported that strain gauges could be used to determine the *in vitro* strain magnitude and distribution within the cement layer indirectly and un-invasively by installation of these gauges on the external surface of crowns.

Goldstein *et al* (1992) simulated the flexion of a metal/ceramic fixed partial denture (FPD) framework under occlusal load. They stated that this flexion could cause fracture of the porcelain or deterioration of the cement seal. This study used a compressive load applied to a four-unit mandibular FPD replacing the second premolar and the first molar. Flexion was recorded with elapsed time holographic interferometry, using 39 porcelain fused-to-metal frameworks cast of silver-palladium alloy. They observed marginal distortion at the distal abutment of the mandibular four units FPD under compressive loading.

Stress relaxation is an important property of materials. However, there is scant information in the dental literature. If a material was strained and this is maintained, the stress within the material decreases with time. The phenomenon of stress relaxation is an indication of elastic behaviour and may be attributed to an elasticity or plasticity. In the latter case stored elastic energy is lost as elastic strain is converted to plastic strain; so that when the applied load is removed the original dimensions of the specimen are not regained leading to permanent deformation. This could lead to loss of adaptation or ditching of the luting material. Paddon and Wilson (1976) defined stress relaxation as the time-dependent change of stress in a material held at a constant total strain. They added that it was closely related to the well-known phenomenon of creep which also was defined as the time dependent change of strain of a material under constant stress. They emphasised that stress relaxation is the exponential time decay of stress within a specimen. Paddon and Wilson (1976) studied the stress relaxation properties of three dental cements, silicate, zinc polycarboxylate and glass ionomer. They found that silicate cement was highly rigid with a high modulus and minimal plastic response. The zinc polycarboxylate cement retained marked plastic properties which decreased to some extent with cement age whilst the glass ionomer cement showed behaviour that was intermediate.

Functional stress on a fixed restoration is resisted by the restorative material, the tooth preparation, and the cement itself; further, the load is greater on the cement lute when the retention form is poor (McLean, 1978).

Smyd (1961) reported that dental cements have high resistance to compressive load of between 6.000 -12,000 pounds p.s.i.. The kinematics of fixed partial dentures sometimes result in stresses applied to only small areas of the dental bond. Pressures rise alarmingly when these small areas bear all the stress of the load. He also reported that brittle materials like dental cement are weak in tension and shear as compared with their strength in compression.

Yettram *et al* (1976) presented stress distributions for an unrestored and a restored mandibular second premolar under masticatory forces. These were obtained using the finite element method of stress analysis applied to two-dimensional models. The effect of the relative stiffness of the materials was examined in each instance. They reported that when two points of occlusal loading were applied to a full veneer crown, the main stresses were distributed to the dentine along the axial walls due the greater stiffness of the crown material compared with the dentine. They also added that the types of stress generated at the cemented joints would be tensile or compressive between the gold crown and the dentine. The type of stress depended on the direction of the load and the relation between the fitting surface of the crown and the dentine surface.

Oilo and Espevik (1978) studied the stress and strain behaviour of some luting cements. Stress/strain diagrams of cylindrical specimens using two different crosshead speeds (2mm/min and 0.1mm/min) at 23°C and 37°C showed that large differences existed between the various luting materials. Zinc phosphate cement exhibited high strength, high modulus of elasticity and a small plastic strain at fracture. Resin cements also had high strength, whilst elastic and plastic strains were also high. A polycarboxylate and an EBA-cement both showed low values of strength and modulus of elasticity combined with a high degree of plastic deformation at fracture. They found that the mechanical properties of the materials were significantly lower when tested at 37°C than at 23°C. They also reported if the biting forces (load) in the molar area were in the region of 300-600N, the stress in the cement film was in the range of 2.5-5 MN/m^2. The stress was not evenly distributed since it depended on the form of the preparation and the direction of the load. They concluded that the stress and strain behaviour of dental cements was affected remarkably by the temperature and load rate.

Kamposiora *et al* (1994) used two dimensional Finite Element Analysis (FEA) to examine the effect of cement type and thickness using a simulated first premolar. They found that zinc phosphate cement exhibited greater stress in the cement layer than other cements with lower elastic moduli. Additionally, they reported that there was minimal difference in stress as a function of cement thickness.

In a further study, Kamposiora *et al* (2000) again used three-dimensional (FEA) of a simulated mandibular first premolar. They investigated the effect of (1) crown margin type, (2) cement type, (3) cement thickness, (4) loading direction, and (5) loading magnitude on stress levels and distributions within the luting cement that might lead to cement microfracture. The crown preparations exhibited shoulder or chamfer margin configurations whilst the cements used were zinc phosphate, zinc polycarboxylate, glass ionomer, and resin in thicknesses of 25 or 100μm. Loads were either axial or oblique at 10 and 100 MPa. Areas and levels of stress concentrations within the cement were determined. They found stresses in the cement were low for all situations except for 100 MPa applied obliquely. Stresses under oblique loads were 10 to 150 times higher than under axial loads. Stresses at the margins of crowns with chamfer margins were higher than those with shoulders. Except for zinc phosphate cement, cement thickness minimally affected stress levels and distribution. Greater stresses were found in the cement with the greater Young's modulus i.e zinc phosphate cement. They concluded that the chamfer margin design could lead to higher stresses than the shoulder margin placing the cement at greater risk of microfracture and possible crown failure. Glass-ionomer and composite resin cements had more favourable mechanical properties for resisting microfracture.

Recently, Proos *et al* (2003) used a similar approach of analysing stress for different cements and cement thicknesses using axisymmetric FEA. They applied an axial load to a first premolar restored with either an all ceramic or metal-ceramic crown. They analysed eight different axisymmetric models using combinations of In-Ceram or gold copings with adhesive resin or zinc phosphate cement as the luting agent at thicknesses of either 50 or 100 μm. They found that the peak tensile stresses in the adhesive resin cement remained below its fracture strength, but exceeded the fracture strength of zinc phosphate cement. They concluded that the influence of the luting agent's elastic modulus on the stresses in the crown was minor and that the influence of luting thickness was even less. It was clear that the role of the luting agent was primarily that of transferring stresses between the relatively stiff coping and the underlying dentin. There was no evidence that the luting agent itself playing a significant role in resisting deflection of the applied force.

Farah and Craig (1974) reported that analysis of internal stresses utilizing the theory of elasticity was difficult because of the complex configuration of dental restorations. As a result, there is a shortage of data relating to this important aspect of cement behaviour.

2.7. Fatigue and dental cements:

Mclean (1979) reported that the dentists must rely not only on the cement lute, but rather on the knowledge of stress analysis and preparation design when constructing fixed restorations. He added that failures in fixed restorations were caused by loss of cement seal either by dissolution or mechanical breakdown.

The discovery of fatigue occurred in 1800s when several investigators in Europe observed that bridge and railroad components were cracking when subjected to repeated loading. As the century progressed and the use of metals expanded with the increasing use of machines, more and more failures of components subjected to repeated loads were recorded (Hoeppner, 1996). Fatigue is so well recognised as a cause of fatal accidents that it is now well established in the vocabulary of the mass media. Under certain loading conditions, a component may suffer a loss in strength over a period of time in service. The component appears to tire. However, it is only within the last 30 years that a reasonable model of fatigue has been developed which can enable a designer to prevent failure happening in service (Smith, 1982).

Baran *et al* (2001) reported that most materials fail in a process termed fatigue when subjected to stress or strain over a period of time. This failure can be manifested as fracture, loss of components or as wear. They added that these were often influenced by environmental factors. Stress or strain could be static (remained constant with time), dynamic (applied at some constant rate), or cyclic (stress or strain magnitude varying with time).

Reid *et al* (1990) also classified two types of load that may cause fatigue symptoms:

1. Reversed (cyclic) loading.
2. Steady loading, in the presence of a chemically active agent which might be as simple as water.

Hoeppner (1996) reported that the fatigue process was difficult to study. Nonetheless, extensive progress on understanding the phases of fatigue has been made in the last 100 years. It is now generally agreed that four phases of fatigue may occur:

1. Nucleation.
2. Structurally dependent crack propagation (often called the “short crack” or “small crack “phases).
3. Crack propagation that is characterised either by linear elastic fracture mechanics, elastic-plastic fracture mechanics or fully plastic fracture mechanics.
4. Final instability.

Torbjörner and Fransson (2004) stated the factors that influence the risk of fractures caused by mechanical fatigue:

1. Magnitude and frequency of the occlusal loads.
2. Direction of forces.
3. Dimension and shapes of dentine and restorative materials.

Lampman *et al* (1996) reported that in fixed prosthodontics technical failures were a result of fatigue fracture. They added that the tooth, cement, and restorative materials are subjected to repeated stresses over a long period of time. When a material, dentine, cement or restorative material is subjected to intermittent tensile stress, a small crack may develop that slowly grows until a fatigue-caused fracture occurs.

Braem *at al* (1994 a) reported that investigators carried out fatigue tests in the laboratory to differentiate between materials and to obtain guidance leading to restorations that are fatigue-resistant.

In the mouth however, loading is often reversible in a chemically active environment. The type of loading that occurs during mastication is compressive with the material underneath the antagonist contact point being surrounded by material with its surface under tension, especially when the restoration is bio-mechanically bonded to tooth structure and when sliding movements during articulation are being considered.

Kovarik *et al* (1992) reported that the use of the fatigue-testing offers advantages. The majority of studies have used tensile testing to measure the fatigue limit but clinically the amount of tensile stress is minimal. Other studies have attempted to mimic the shear and compressive forces that are found clinically but have done so by applying one catastrophic blow to failure. This mode of failure would occur only for traumatic injuries. Sub-critical load applied over many cycles more closely models the actual service requirements. The authors used 34 teeth to determine the durability of three different core materials beneath crowns in endodontically treated teeth. They used an axial load of 75 pounds which was applied over one million cycles. They stated that 75 pounds was considered to be at the high end of the typical forces exerted during parafunctional activity. They employed a transducer to detect the horizontal and vertical movement of the cast crowns in relation to the tooth. The transducers consisted of a strip of beryllium-copper alloy formed into a box with the ends welded to form the sensing tip. They chose arbitrarily a horizontal movement of 30μm or more as being indicative of the failure point of the casting which represented the break of the cement seal. The choice of the 30µm end point was based on the recommendation of Outhwaite *et al* (1982) who arrived at this figure in a test carried out by a group of restorative clinicians; they evaluated various sizes of gaps between restorations and tooth structure using explorers on extracted teeth. The clinicians reported that they would replace restorations containing gaps or steps of more than 30µm at the interface between tooth and restoration.

Kovarik *et al* (1992) found about 50% of the failures occurred as a result of failure of the zinc phosphate cement at the crown-core interface, the post-root interface or both. Additionally the high number of cement failures and the post fractures for composite resin cores was related to the material's low modulus of elasticity and they claimed that this was due to the more significant shearing stresses on the cement interface.

Recently Proos *et al* (2003) however indicated that the role of luting agent's elastic modulus on stresses in the crown was minor. They stated that the role of luting agents was to transfer the resulting stress between the relatively stiff copings and the underlying dentine.

2.8. Fatigue in restorative dentistry:

In the Academy of Prosthodontics, Glossary of Prosthodontic Terms (2005), fatigue was defined as the breaking or fracturing of a material caused by repeated cyclic or applied load below its yield limit; usually viewed as minute cracks followed by tearing and rupture; also termed as brittle failure or fracture. Manson and Hertzberg (1973) defined fatigue life as the number of cycles a material will withstand before it fails.

Braem *et al* (1994) reported that there are numerous papers describing *in vivo* and *in vitro* fatigue studies with little consideration of the fatigue process itself. Huysmans and van der Varst (1995) emphasised that fatigue-testing is relatively rare in dental research and has been used almost exclusively with restorative materials. Testing conditions with restorative materials have generally differed too much for direct comparisons to be made, but in some studies some conclusions have been drawn. They reported that chewing and swallowing, and parafunctions such as clenching and bruxing cause cyclic stress patterns in teeth and restorations. This is referred to as "cyclic loading". In combination with other factors such as thermal and chemical influences, cyclic mechanical loading causes a large proportion of the failures of all dental restorations. Mechanical properties of dental materials and restorations are usually determined using quasi-static tests, where the load is applied and increased until failure occurs. However, the value of the initial mechanical behaviour for prediction of long-term clinical behaviour is questionable. Cyclic loading causes the material to fatigue and the speed of deterioration may differ from one material to another.

2.8.1. Fatigue sequence and its complications:

Taggart (1907) compared cement grains with a pile of sand. When pressure is applied, those which move out of the way do so, but the others remain packed on top of each other and the greater the pressure the less their ability to assume a new position.

Acid-based cements like zinc phosphates are non-adhesive and act by increasing friction between the tooth and the restoration through micro-mechanical retention. The cements enhance this frictional fit by filling the minor irregularities between the castings and the axial walls of the tooth preparation, so the frictional strength of the cement is the key factor in retention of the crown (Noonan and Goldfogel 1991).

Wiskott *et al* (1995) described the fatigue failure as the development of microscopic cracks in an area of stress concentration. Upon load continuation these cracks fuse to produce an over-growing fissure that insidiously weakens the restoration. Catastrophic failure results from a final cycle that exceeds the mechanical capacity of the remaining sound portion of the material. They reported that fatigue failure is initiated by microscopic cracks that develop in areas of stress concentration. The most common local stress concentrations are grain boundaries, inclusions, local intrusions and sudden changes in the geometric configuration of the surface.
The initial step of nucleation represents a mandatory stage of fatigue failure. When a fissure has reached its critical size, it will definitely progress at each loading cycle. This progress is propagation which represents about 90% of the fatigue life.

They reported that functional loading of teeth implies a multidirectional force pattern that comprises compressive as well as bucco-lingual components. Loading a prosthodontic test structure uniaxially will thus only partly reproduce the mechanical conditions of the oral environment. For descriptive purpose, they reported that as fatigue processes are determined by a progressing fissure in materials that present some degree of ductility, the crack front may leave a groove in the walls of the crack at each loading cycle. A typical pattern of ripples will develop that are visible using electron microscopy: these are termed fatigue striations.

Huysmans and van der Varst (1995) reported that during fatigue cycling, the damage which is initially present in a material gradually grows and decreases the load-bearing capacity of a structure. The structure behaves as if subjected to an ever-increasing load, and it is the effective load rather than actual load that determines the failure probability.

Kamposiora *et al* (1994) proposed that long-term survival of a cemented prosthesis depends on the integrity of the cement layer. Cement fracture permits microleakage of bacteria or their by-products that would change the stress distribution in supporting tissues. They reported that there are many factors that contribute to pulp necrosis after tooth preparation, bacterial microleakage is one important factor among them.

2.8.2. Methodology for fatigue study:

Occlusal forces found in the mouth are repetitive and generally light. DeBoever and McCall (1978) reported that masticatory forces seldom exceed 10-15 pounds. Huysmans *et al* (1992) emphasised that fatigue studies are more informative than single load impact studies, since in the oral environment, the forces applied are more likely to be of cyclic nature and well below the ultimate strength of the foundation restoration. Therefore, cyclic loading mimics more accurately the physiological conditions of mastication. Further, they reported that there are multiple methodologies during testing procedures and analysis of results. The progress in materials and prosthodontics research required a standardisation of the procedural aspects of fatigue testing.

Wiskott *et al* (1995) stated guidelines for fatigue testing of dental materials as follows:

- The baseline for fatigue testing is the fracture mechanism in which dental researchers need special training.
- The clinically relevant number of cycles is 10^6 cycles.
- They recommended the negative stress ratio for prosthodontic structure.
- The whole S-N diagrams are not required as the low cycle regime ($<10^6$ cycles) is applicable only to temporary materials and structures.

- The staircase analysis was recommended to determine S_n.(conventional endurance limit at number of cycles in a fatigue test) and should be used with greater than 20 specimens.
- More cycle numbers of large sample sizes require great cycling frequencies. Such accelerated test methods may require mathematical methods to relate such data to the relevant clinical frequencies however.

2.8.3. Problems with fatigue testing:

Reid *et al* (1990) reported that the intra-oral degradation of restorative materials is a complex process and has not been mimicked to any great extent by simple laboratory tests. Each group of materials such as metals, polymers and cements seems to fail by mechanisms specific to that group and generalizations are difficult. Evaluation of the long-term behaviour of restorations in clinical trials can be time-consuming and difficult, and that an alternative might lie in fatigue testing.

Failure of dental restorations may occur at any stage; early failures may be related to procedural mistakes or material flaws. Failure after an extended period of time is more common and can be attributed to many environmental factors. One of these is the cyclic loading to which restorations are subjected during chewing and parafunctional activities.

To determine the life expectancy of a material in a controlled clinical trial is desirable. However, many trials require long periods of time and a large number of patients to collect sufficient data. A partial alternative can be found in an *in vitro* investigation into the influence of cyclic loading on these materials (Huysmans *et al*, 1993).

Huysmans and van der Varst (1995) stated that few clinical studies have validity because of their retrospective nature. Thus, few experimental data are available on long-term behaviour or life expectancy. *In vitro* studies have been performed to provide some information but their limitations are obvious. In a clinical situation, initial failures are rare and restorations are expected to function for a number of years. Life-expectancy, however, depends on several factors. Cyclic mechanical loading as a result of occlusion and articulation contacts is an important factor, since it causes a gradual reduction in strength, termed fatigue. Cyclic loading can be introduced into an *in vitro* experiment. By increasing the frequency of loading and by omission of the periods of non-loading which occur clinically, an accelerated mechanical deterioration process can be affected.

2.8.4. Techniques:

2.8.4.1. Staircase method:

A first step toward more efficient testing was the staircase method (Dixon and Mood, 1948). This method was used to establish a censoring point for a chosen lifetime, the cyclic load for the probability of failure at a later time, or $P_{failure}$=50%. By raising the load for the next specimen if a specimen survives, and lowering the load if it fails, the staircase method concentrates testing around $P_{failure}$=50%.

If at a given stress level the sample does not fail within 10000 cycles, the stress level for the next sample is increased by 4% of the restrained fracture strength. If failure occurs, the stress level is decreased accordingly. The number of cycles is arbitrarily determined, based on pilot testing (Braem *et al,* 1995 & Braem *et al*, 1994).

Draughn (1979) described the staircase method as an "up and down" method. Tests are conducted sequentially with maximum applied stress in each succeeding test being increased or decreased by a fixed amount, according to whether the previous stress resulted in a failure or no failure.

2.8.4.2. Boundary technique (Maenning, 1970):

The Boundary technique is another test method. This technique was cited by (Huysmans *et al*, 1992). The reference was written in Dutch language. The boundary method initially follows a staircase strategy. Later, two load levels are chosen at which specimens are tested. Assuming a normal strength distribution, the 50% failure load is found by interpolation between these two levels.

A censoring point (a fixed number of cycles) is selected at which the test is broken off and the specimen inspected. To find an appropriate load, consecutive tests are performed at levels that are raised after each test by a predetermined interval, until the first failure occurs at or before the censoring point is observed. This load lies with certainty within the range of transition (the load range between almost certain failure and almost certain survival) for the censoring point chosen. At this level, several specimens are tested. The probability of failure is then calculated from the number of failures using a specific formula.

2.8.4.3. Preliminary failure:

Libman and Nicholls (1995) defined the preliminary failure as the formation or propagation of a crack in or around the crown luting cement layer as a result of cyclic loading. Libman and Nicholls (1995), Fan *et al* (1995), and Freeman *et al* (1998) applied fatigue loading to endodontically treated teeth to determine the number of cycles needed to create cement failure. This type of cement failure was termed preliminary failure, in that while the cement had failed, integrity of the restoration was maintained. Clinically, this type of failure results in microleakage and can potentially cause recurrent caries, loss of crown retention, post fracture or complete dislodgement of the crown and foundation restoration.

Reisbick and Shillingburg (1975) used zinc phosphate cement in their study to investigate the effect of geometry in the resistance and retention of gold cast restorations. They fatigued one half of the samples from each classification for 8 hours, 48 hours after cementation. The fatigue load was directed horizontally and alternated between 0-60 pounds for 1000 cycles. The crowns in both groups were then subjected to an increasing tensile load until failure. They reported that there was no statistical difference between the fatigued and non-fatigued samples. They recommended that greater loads or longer non-axial loading cycling times should be used in fatigue studies. They added that the common mode of failure of zinc phosphate cement is cohesive with cement partially retained on the casting and preparation. Nicholls (1974) reported the fact that cement can be found on both the crown and tooth after failure which led to the conclusion that failure occurs within the cement layer itself. This presupposed that the stresses induced in the cement layer caused fracture which ultimately led to loss of retention. He added that the objective of any investigation relating to retention is to determine these stresses and if possible to determine if an optimum geometry could be found that minimised them.

2.8.5. Experiments in literature:

Huysmans *et al* (1993) evaluated the failure behaviour of teeth restored with post and cores when subjected to cyclic mechanical loading and compare this with quasi-static failure.

The specimens were subjected to cyclic loading with frequency of 5Hz at an angle of 45°. The load levels were 50, 60, 65 and 70% of the mean quasi-static failure loads. The number of cycles was greater than 10^5 with the maximum testing time set at 10^6 cycles.

However, extra evaluations were included after 10^4 and 10^5 load cycles, giving a total of three evaluations. The specimens were checked visually after 10^4 load cycles. If failure had not occurred at that time, they were remounted again in the testing machine and loaded until the next evaluation at 10^5 load cycles.

The criterion of failure is related to the time of functioning: the specimens were divided into three groups, short life (failure before 10^4 cycles), intermediate life (surviving at 10^4 but failure before 10^5 cycles) and long (surviving at 10^5 or more).

The nature of their study was descriptive and qualitative whilst the small numbers of specimens for some failure types precluded statistical analysis of the data.

Kovarik *et al* (1992) used 75 pounds for one million cycles and they claimed that was the highest end of the average of typical force that patient could generate during para-functional activities. Outhwaite *et al* (1982) reported that one million cycles was equal to 5 years of heavy wear in the mouth.

Cohen *et al* (1997) used cyclic fatigue for testing endodontic posts. 22.2 N (5Pounds) were used for 4 million cycles. This number of cycles would be equivalent to 10 years of clinical service. Their work was based on that of

Huysmans and van der Vast (1995) who believed that the best model would be equivalent to 5 millions load cycles. They stated that this amount of cycles would be equivalent to 5-15 years of clinical service.

Huysmans and van der Varst (1995) used cyclic loading to examine the long-term mechanical behaviour of premolar teeth restored with a titanium alloy post and an amalgam or composite core. They used a frequency of 5 Hz at an angle of 45° to the long axis of the tooth. The crowns were immersed in Ringer's solution at a temperature of 32° simulating the oral environment. The boundary technique was used for determination of the mean fatigue strengths of the restorations at 10^4,10^5,10^6 cycles, simulating up to 1-3 years of clinical function.

Gateau *et al* (1999) investigated post and core restorations under artificial crowns using fatigue testing with a load of 400N for 1.5 million cycles.

Fan *et al* (1995) investigated different treatment modalities for rebuilding structurally compromised maxillary first premolars. The modalities were: (A) buccal stainless steel Parapost/amalgam core, (B) palatal stainless steel Parapost/amalgam core, (C) two stainless steel Paraposts/amalgam core, (D) two regular Link Plus TMS single-shear Minim pins/amalgam core, and (E) palatal cast gold post and core.

They used a load fatigue of 5.2kg (51N) with frequency of 72 cycles/minute for 250000 cycles. To detect the preliminary failure they used strain gauges as adjuncts which were cemented across the crown margins to monitor the movement of the crown relative to the tooth. The load used in this study (51N) was claimed to be comparable with the occlusal functional force. The orientation of loading was at 135° to the long axis of the tooth against the palatal side of the buccal cusp. Each specimen was placed in a room temperature water path during testing. The relative movement of the cemented crown on the tooth was based on the amplitude of the strain. Following preliminary failure, each specimen was immersed in a black ink for 12 hours to demonstrate any of microleakage.

They found that the buccal Parapost/amalgam core and the two Minim pins/amalgam core specimens failed prior to the upper limit of 250000 load cycles. Additionally, both the palatal post/amalgam and palatal cast gold post and core reached the upper limit of 250000 load cycles without preliminary failure.

Junge *et al* (1998) compared the number of cycles to failure of 15 human central incisors restored with full cast crowns and cemented with 3 different luting cements. They applied a fatigue load of 1.5 kg at a rate of 72 cycles/minute. They added that once a crack forms in the cement layer, micro-movement of the cast crown relative to the finish line which is not discernible to the unaided eye can be observed through the output of the strain gauge on the lingual tooth surface.

Following preliminary failure, each crown was sectioned and visually inspected under 10X magnification. The examination was to reveal whether the cement adhered to the tooth or crown or both. Three types of cement fracture location were defined, (1) tooth-cement fracture (TC), (2) Crown-cement fracture (CC), and (3) Mixed fracture (M).

Kamposiora *et al* (1994) used finite element analysis to study stress in cemented gold crowns with either a shoulder or chamfer finish line. The stress at the midpoint of the axial walls of crowns cemented with zinc phosphate cement was 2-3 times greater than with other cements. The stress patterns indicated that shear strength appeared to be predominant and that these stresses would be magnified on teeth with short and/or over-tapered axial walls. Consequently, zinc phosphate cement may be less desirable in such situations as its tensile strength is lower than polycarboxylate, glass ionomer and resinous cements.

Freeman *et al* (1998) applied cyclic loading to central incisors that were endodontically treated and restored with a post and core and cast crown. Restored teeth were fatigue-loaded until preliminary failure of the casting occurred as detected by a strain gauge bonded across the lingual margin of the cast crown. After preliminary failure, fatigue loading was continued for 100,000 load cycles with the crown margin exposed to basic fuschin dye. Teeth were then immersed in dye for 24 h, sectioned, and evaluated for leakage. There was no significant difference in the number of load cycles required to cause preliminary failure among the three post and core systems. Leakage occurred in all three groups, with no significant difference between groups. They concluded that the occurrence of preliminary failure is clinically undetectable, yet it allows leakage between the restoration and tooth that may extend down the prepared post space.

2.8.6. Fatigue test simulation:

Renggli *et al* (1983) reported that the average time of tooth contact *per* day, due to chewing and swallowing, was estimated at *ca.*18 min. Bates *et al* (1975) reported that chewing rates vary between 60-180 strokes *per* day. Therefore, 10^6 cycles would represent *ca.* 1-3 years function.

Huysmans and van der Varst (1995) reported that determination of the relationship between the number of loading cycles and strength is a first step towards making better prediction and estimates of clinical longevity.

Wiskott *et al* (1996) investigated the relationship between taper and resistance of cemented crown with different luting cements subjected to dynamic loading. They applied rotational dynamic stress to the samples until the bond failed or the components reached 10^6. The data were analysed using staircase technique and were reported in section 2.2.1.of this review.

Huysmans and van der Varst (1995) reported that 400N simulating the higher muscular force produced during functional and para-functional activities. They claimed that the cyclic conditions were estimated to be equivalent to 4 years of normal mastication.

400N has been used by many workers as it represents the value for average biting force (Floystrand *et al*, 1982), (Bakke *et al*, 1990), (Waltimo *et al*, 1994), (Mericske-Stern *et al*, 1995), (Waltimo and Kononen ,1993), (Braun *et al*,1995), (Braun *et al*, 1996), (Aydin and Tekkaya ,1992), (Erhardson *et al*, 1993), (Gateau *et al*, 1999).

Gundler *et al* (1993) investigated crown retention with varying angle of convergence cemented with zinc phosphate cement on human teeth after fatigue cycling. They were loaded in a range between (50N-700N) at the frequency was 2Hz. The test continued until the crown loosened or until a maximum of one million cycles had been achieved. They found no crowns loosened at a taper of 20°, but at 40° 7/10 crowns loosened. All crowns with a 60° taper loosened.

Wiskott *et al* (1999) reported that film thickness is a significant factor in resistance to fatigue failure. Under cyclic loading, thinner cement interfaces have higher failure values. They indicated that this behaviour is known, but ill-defined in fatigue research; they supported their explanation by Griffith's theory of flaws.

Griffith's theory of flaws (Cotterel and Mai, 1996):

1. All materials contain defects that are randomly distributed inside the material.
2. Under load, each of these defects is likely to initiate crack growth.

3. The higher the pile of the material, the more numerous the defects and the larger the probability of crack initiation.

2.9. Conclusions of review of literature:

1. Geometry of the preparation affects retention and resistance of a cemented crown.
2. Different cements behave differently under both mechanical and solubility testing.
3. No one *in vitro* test can replicate the conditions which affect the durability of a crown in clinical service.
4. *In vitro* tests may predict aspects of the behavior of a cemented crown *in vivo*.
5. Recent *in vitro* studies have included a more frequent use of dynamic testing.
6. The literature contains little evidence on the behavior of the cement film beneath a crown when submitted to dynamic loading.
7. The majority of the investigations were tensile in nature with scant information about the compressive loading.
8. Using the tensile tests investigate only the tensile or shear strength of the luting cement omitting the compressive strength.

Chapter 3: Statement of the problem

There is little or no clinical information on the performance of luting cements beneath full crowns. This is not surprising given the variables and long duration required for such studies.

Information derived from *in vitro* work has come, in the main, from tensile testing or non-axially directed forces. The former is recognised as unrealistic, whilst the latter is complex and the data are difficult to interpret. In addition, the information derived regarding the behaviour of the cement itself is limited. There were little data which provided useful information about the response of the complex made up of the preparation, the crown and the cement lute.

It was decided that this would benefit from further investigation using an *in vitro* model. It was also considered important that the study monitored the behaviour of the cemented crown during mechanical testing and that the loads applied to the cemented crown should be dynamic. The latter was considered important as being more clinically representative than static loading.

Most previous work had tested cement films to the point of clinically detectable failure of the crown. It was considered relevant to undertake a study in which any deterioration of the bond between the preparation and the crown might be monitored over time.

Chapter 4: Aims and objectives

4.1. Aims

The aim of this investigation was to examine *in vitro* the behaviour of the cement film used to retain a full crown on its preparation.

4.2. Objectives

The study had the following objectives:

1. To construct an appropriate series of dies and crowns.
2. To measure the strain in the axial walls of the crown when subjected to occlusal loads.
3. To relate the values for strain to the presence and condition of the underlying cement film.

4.3. Null Hypothesis

The durability of the attachment of a cast full gold crown to a die was not influenced by the type and magnitude of the applied load, or variations in the geometry of the die or immersion in different media.

Chapter 5: Programme of work

The first part of this work was fabrication of appropriate experimental models. This consisted of:

1. Construction of metal dies of a full crown preparation which allowed repeated use and recovery of full gold crowns cemented to them.
2. Construction of a replica of a full gold crown preparation which was suitable for modification.
3. Fabrication of full gold crowns using standardised techniques which permitted the installation of strain gauges on their axial walls.

The following investigations were made of the strains under axial loading in the axial walls of the crowns on their respective dies.

(1) Under static loads:

- 1.1. The effect of cement distribution.
- 1.2. The effect of cement thickness.
- 1.3. The effect of the geometry of the crown preparation (taper and axial height).
- 1.4. The effect of variables of the crown itself (increase occlusal surface thickness and heat treatment).

(2) Under dynamic loads:

It included the effect of hydration and various aqueous media.

2.1. The effect of dynamic loading on cemented crowns under dry condition (Control).

2.2. The effect of dynamic loading on cemented crowns immersed in distilled water.

2.3. The effect of dynamic loading on cemented crowns immersed in lactic acid during day and distilled water during night.

2.4. The effect of dynamic loading on cemented crowns immersed in lactic acid during day and night.

(3) Scanning electron microscope investigation of the dynamic loaded samples.

Chapter 6: Materials and methods

This work examined the behaviour of the cement film between a cast full gold crown and a nickel chromium die. The protocol was designed to produce a model that could be used more than once and allowed a number of identical dies and crowns to be produced.

Various materials could have been used to form the die:

1. Extracted human teeth.
2. Bovine dentine.
3. Resins.
4. Metal.

6.1. Model and die construction:

6.1.1. Master model fabrication:

A section of parallel sided plastic rod of diameter 11mm was selected. A plastic container which fitted over the plastic rod was chosen as an impression tray and an impression was made using vinyl polysiloxane impression material (Coltene/Whaledent AG. Feldwiesnstrasse 20, 9450 Altstatten/ Switzerland). Once set, the impression was separated from the rod and left for more than 24 hours (Figure 6.1).

Using Type IV dental stone (MOLDASTONE®, Heraeus Kulzer Gmbh&co.KG, Germany), the impression was cast and the resultant die used as the master model. This was left to harden for twenty-four hours before being mounted axially on the horizontal table of a surveyor (Krupp Dental, Dentagraph, freid-krupp GmBH,Essen,Germany) and retained by a low expansion articulating plaster (Shera, workstoff-technologie,GmbH&co.KG, Germany) (Figure 6.2). The model was milled in a type 990 KaVo milling machine using a 6° tapered bur (Figure 6.3) in straight hand piece to produce a full crown preparation with 12° total occlusal convergence (TOC). The approximate dimensions of the preparation were axial wall height 6mm and base diameter 11mm, with a finish line width of 0.5mm. The finishing line was a chamfer and the occlusal surface flat.

6.1.2. Metal die fabrication:

An impression of the prepared stone master model was made using vinyl polysiloxane impression material. The impression was left for more than twenty-four hours, removed from the tray and sectioned longitudinally using a scalpel blade to give two halves.

The two halves were reassembled and returned to the impression tray. A hard blue inlay casting wax® (Kerr, USA, 28200 Wick Road, Romulus, MI 48174-2600) was melted incrementally into the impression and left to bench cool for about 45 minutes, the impression was then removed from its tray and the wax pattern released from the impression. Fifteen wax patterns were produced. Each pattern was sprued and mounted in a X3 casting ring lined with one layer of cellulose ring liner. The pattern was invested using a phosphate-bonded investment (Hervest® Speed E, Heraeus Kulzer Laboratory Products Division, Germany) under vacuum at the manufacturer's recommended powder:liquid ration of 180 g of powder to 37 mls of special liquid and distilled water. The investment was allowed to harden for one hour before being placed in an oven at 900°C for about an hour for burn-out and heat-soaking; it was then cast in a nickel-chromium alloy (Heraeus Kulzer Laboratory Products Division, Germany). The castings were left to bench cool before being divested and cleaned by sand-blasting with aluminium oxide powder (Fino GmbH, DT Bad Kissingen, Mangelsfeild 11-15, D-97708 Bad Bocklet, Germany). The sprue was removed and the casting finished and polished.

The base of each die was cut in a standardised manner using a lathe-mounted tungsten carbide tip to provide a base perpendicular to the long axis. The die was perforated at the centre of its base using a milling machine with a cobalt drill of 2.5mm diameter (HSC Stub Drills, RS Company, UK). The resultant hole was then tapped with a m34 spiral point (Blue, Spiral point-E344&Blue, Spiral Flute-E346 Taps, UK). A stainless steel screw was fitted into the tapped hole. A total of 23 master nickel-chromium dies were produced each with a total occlusal convergence (TOC) of 12° and axial wall height of 6mm. Each die was numbered.

Modifications were made to the stone master model. It was further prepared to produce a model with 24° total occlusal convergence and 6mm axial wall height (Figure 6.4). It was then further re-prepared to produce a TOC of 24° and 8mm axial wall height.

An impression was made of the stone master model after each re-preparation, and cast nickel-chromium dies were made as previously described to create the following:

a) 5 nickel-chromium dies with 24° TOC and 6mm axial height.

b) 5 nickel-chromium dies with 24° TOC and 8mm axial height.

This produced the following:

- 23 dies (12° TOC and 6mm axial height).
- 5 dies (24° TOC and 6mm axial height).
- 5 dies (24° TOC and 8mm axial height).

Figure 6.5 showed one die of each specification.

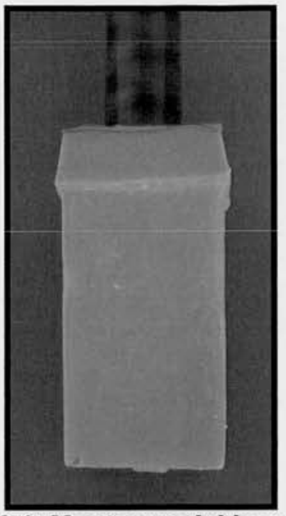


Figure 6.1: Master model impression

Figure 6.2: Master model on surveyor platform

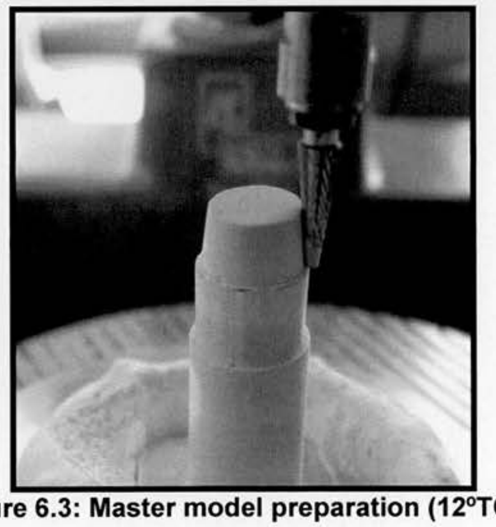


Figure 6.3: Master model preparation (12°TOC)

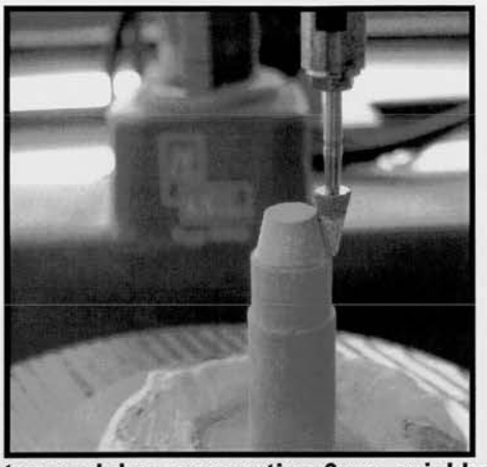


Figure 6.4: Master model re-preparation.6mm axial height (24° TOC)

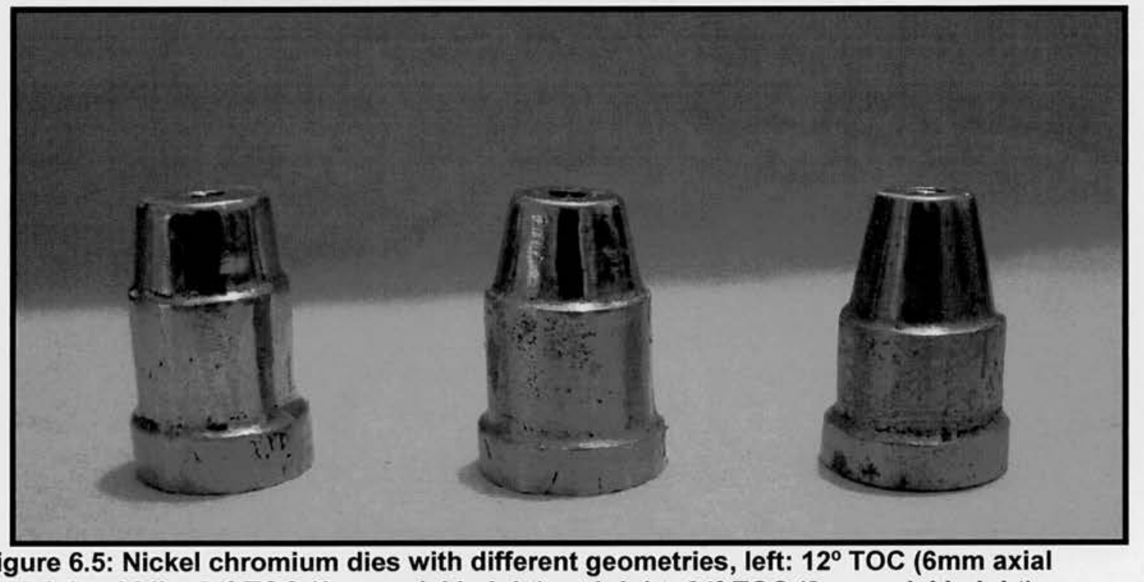


Figure 6.5: Nickel chromium dies with different geometries, left: 12° TOC (6mm axial height), middle: 24° TOC (6mm axial height) and right: 24° TOC (8mm axial height). (From left to right)


Figure 6.6: Nickel chromium master die (12° TOC) finished, polished and coated with die spacer

6.2. Gold crown fabrication

Two layers of die spacer (TRU-FIT, Geo.TAUB PROD. &FUSION CO.INC.Jersey City, NJ, U.S.A.) were painted on the metal die to within 1mm of the margins and allowed to dry for 10 minutes (Figure 6.6). Separating medium (Microfilm, Kerr, Italy S.P.A) was applied, thinned with compressed air and allowed to dry for 5 minutes. A wax pattern was fabricated by means of a wax dip technique (Bellwax ™ ,Kerr USA). The die with its central screw in place was dipped in the molten wax at 70°C for 6-8 seconds covering the entire preparation's axial surface and extending up to 2mm beyond the preparation finish line. The wax was left to harden, carved to produce a smooth surface, then removed from the die and the thickness verified using a wax gauge calliper (Iwanson Type Measuring) at multiple points over its surface. The occlusal surface of the pattern was carved flat with the thickness of the pattern adjusted until about 0.5mm thickness was reached. The pattern was replaced on the die and the margins re-melted and finished (Figure 6.7).

The pattern was sprued, mounted on a former, coated with surface wetting agent (Hera SWE 2000, Germany) and invested (Hervest® Speed E, Heraeus Kulzer Laboratory Products Division, Germany) in an X1 casting ring lined with cellulose. The mixing of the investment was under vacuum and followed the manufacturer's recommendation. For each wax pattern 60 g of powder and 13 mls liquid were mixed.

The initial investment material went out of production after the first 15 dies were made; therefore a second material was used for the gold crowns in later experiments. This was a graphite-free, phosphate-bonded investment material (MOLDAVEST®futura). The powder liquid ratio was 60 gram of powder to 13mm of liquid.

The mix was first stirred by hand until the powder was thoroughly wetted, then placed in a vacuum mixer (Refer, Twister Pro, Germany) for 15 seconds under vacuum but without spatulation, then for a further 60 seconds being mixed under vacuum at room temperature.

The ring was filled with the investment under vibration. 20 minutes after setting, the top of the investment was scraped and the casting ring transferred to a furnace (KaVo burnout furnace, type 5636, Germany) at 700°C for 30 minutes prior to casting in an induction casting machine (Heraeus Kulzer, Heracast IQ, Germany). The casting was made in type IV yellow gold alloy (Bodent 60, Charles Booth, 49-63 Spencer Street, Birmingham, B18 6DE).

The casting was quenched in cold water before divesting. Final removal of investment was accomplished by air abrasion with 50μ aluminium oxide powder at a pressure of 5 bars. The sprue was removed using a separating disk and the gold crown was finished (Figure 6.8) using brown and pink stones and then polished on the die using a green rubber wheel. The adaptation and accuracy of fit of the crown were verified. The crown was marked using a tungsten carbide bur (Jet Carbide, Beavers Dental, Morrisburg, Ontario K0C 1XO, Canada) in a friction grip hand piece to scribe a thin line from the gold crown onto the nickel-chromium die to allow repositioning. In addition, a black marker was used to define the margin of the crown.

The gold thickness was checked again at four random locations axially and occlusally using a metal calliper gauge (Iwanson Type Measuring). The final fit of each crown was visually verified at X4 magnification. A total of forty three gold crowns was made and were divided into groups for testing.

Forty three crowns were produced, according to the test protocol.


Figure 6.7: Wax pattern on finished die.

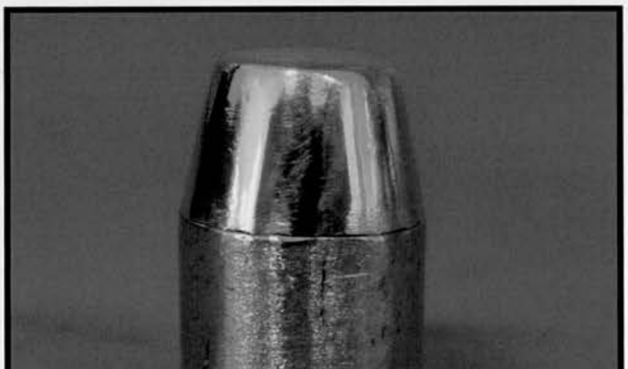


Figure 6.8: polished gold crown on its die.

6.3. Strain gauge installation

6.3.1. Strain gauge specification

The type of strain gauge used was EA-06-031EC-350 (Vishay Measurements Group UK Ltd, Stroudley road, Basingstoke, Hants RG24 8FW, UK) which was a constant strain gauge widely used in experimental stress analysis. The strain gauge had a resistance of 350Ω (Ohm) +/-0.2% at 24°C (Figure 6.9).

The strain gauge was of open-faced construction with a 1mil (0.0254mm) tough, flexible polyamide film backing and a strain limit of approximately 3%. Each strain gauge had two soldering terminals (Bondable Terminals, CPF-50C, Measurements group, Inc., Raleigh, NC). The dimensions of the strain gauges used in this study were recorded in Table 6.1.

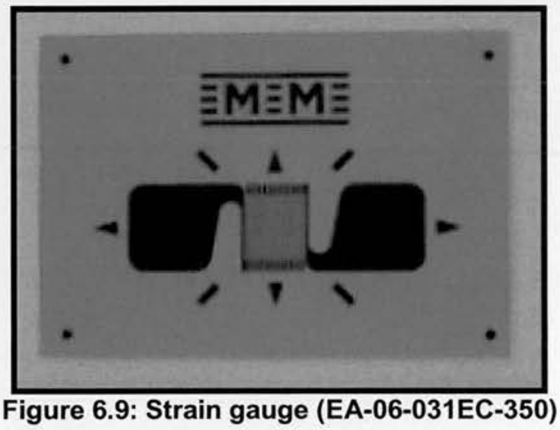


Figure 6.9: Strain gauge (EA-06-031EC-350)

Gauge Length	Overall Length	Grid Width	Overall Width	Matrix Length	Matrix Width
0.79mm	1.07mm	0.81mm	3.56mm	4.3mm	5.8mm

Table 6.1: Dimensions of strain gauge (EA-06-031EC-350)

6.3.2. Strain gauge placement

The strain gauges were placed on the axial surfaces of the gold crowns; two strain gauges were installed opposing each other, about 1mm above the gold crown margin, perpendicular to the long axis of the crown-die assembly.

Installation started with a thorough degrease of the gauging area with isopropyl alcohol (IPA) using a clean gauze sponge. Further, the surface of the crown was lightly abraded with 220-320-grit silicone-carbide paper to remove any surface oxide. Final preparation was made with 320-grit silicone-carbide paper on surfaces thoroughly wetted with *M-Prep* Conditioner A (Vishay Measurements Group UK Ltd, Stroudley Road, Basingstoke, Hants RG24 8FW, UK); this was followed by wiping with a dry clean gauze sponge. The residue of the conditioner was removed by slowly wiping with a gauze sponge. A liberal amount of *M-Prep* neutraliser 5A (Vishay Measurements Group UK Ltd) was then applied and the surface scrubbed with a cotton-tipped applicator. The surface was once again dried with a single, slow wiping motion of a gauze sponge. The gold crown was placed on the nickel-chromium die to stabilise it and assist in the strain gauge placement.

The gauge was removed from its mylar envelope with tweezers avoiding touching any exposed foil. The gauge-bonding side was placed on a clean glass plate and the gauge removed by peeling the backing tape at a shallow angle (about 30^{o}) before transferring it to the gold crown.

The gauge was positioned at the prepared site with its tape partially peeled back so that the gauge still remained attached. The back of the gauge and the surface of the crown were coated with a thin layer of adhesive catalyst M-Bond 200 (Vishay Measurements Group UK Ltd) which was allowed to dry for one minute.

The tucked-under tape on the gauge was lifted with the gauge held in a fixed position. 1-2 drops of adhesive were applied at the junction of the tape and crown surface ensuring coverage of the gauge area and its boundaries. The strain gauge was held in position using finger pressure for a minute to allow the adhesive to harden. The tape was then carefully peeled off the gauge completely. The area around the gauge was cleaned using IPA solvent, then left to air dry.

6.3.3. Strain gauge soldering technique

Three 1.0 metre length wires were soldered to the terminal of each strain gauge (10/0.1 Miniature wire; Farnell InOne, Canal Road, Leeds, LS12 2TU). The installed gauge required a resistance to ground of at least 10000-20000 megohms. The leakage resistance was checked with a Model 1300 Gauge Installation Tester (Vishay Instrument, 63 Lincoln Highway, Malvern, PA 19355-2143).

Two gauges were attached to each crown as shown in Figure 6.10.

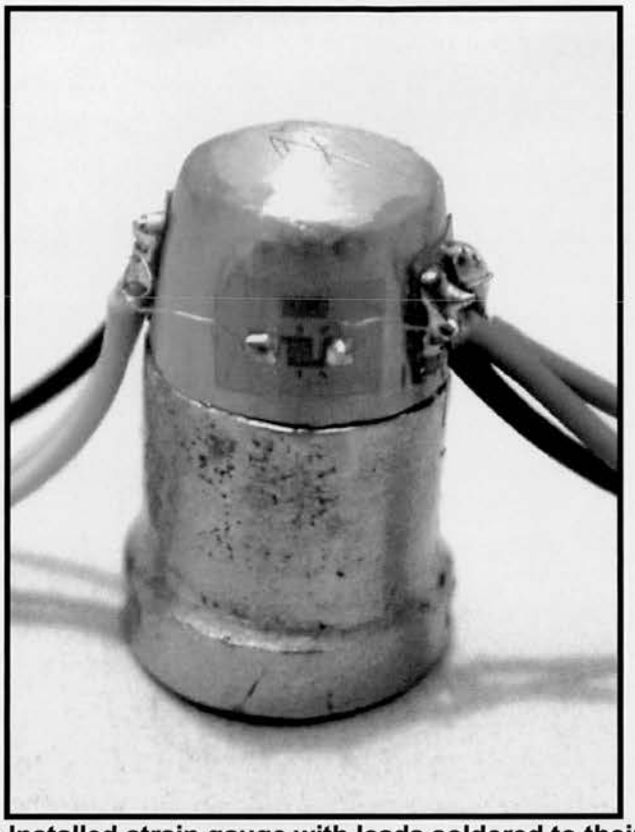


Figure 6.10: Installed strain gauge with leads soldered to their terminals.

6.3.4. Three-Wire Circuit connection

The commercial static strain indicators employed a form of the Wheatstone bridge circuit to detect resistance change in the gauge with strain. Lead-wire effects were virtually eliminated in these single active gauge installations by use of the "three-wire" circuit (Figure 6.11). In this case the third lead, representing the centre point connection of the bridge circuit, was brought out to one of the gauge terminals.

For this method of lead wire compensation to be effective, the two lead-wires in the adjacent bridge arms were of the same length and were maintained at the same temperature. The three-wire circuit was the standard method of connection for a single active temperature compensated strain gauge in a quarter-bridge arrangement.

Each strain gauge was connected with wire of a different colour from its opposing one to facilitate their connection to the amplifiers. In this and successive experiments the same lead was connected the same amplifier (Strain Gauge Amplifier EPS HWU) to record strain on the axial walls of the gold crown as digital figures.

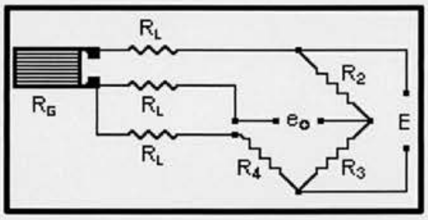


Figure 6.11: Three-wire circuit for single active gauge (Quarter Bridge)

6.4. Crown cementation

Before the cementation procedure, the central screw was replaced in the die flush with its occlusal surface. The alignment between the crown and the die was verified using the line scribed on their external surfaces.

The cement used for all cementations was zinc phosphate cement (De Trey®Zinc, DENTSPLY Limited, Hamm Moor Lane, Addlestone, Weybridge, Surrey KT15 2SE). The cement met the requirements of the ISO specifications of 9917:1991 for water-based dental cements where the powder: liquid ratio is 2.8g: 1g.

A series of cementations was made. The cement used in the first experiment (static loading) was from a different batch from that used in later testing (dynamic loading).

The batch and lot numbers of both cements were displayed in Tables 6.2 and Tables 6.3.

Property	Powder	Acid
Batch No.	105018/0	105019/0
Lot No.	0207000178	0005000724

Table 6.2: Zinc phosphate cement for (static) loading experiments

Property	Powder	Acid
Batch No.	105018/0	105019/0
Lot No.	0405001853	0407002196

Table 6.3: Zinc phosphate cement for (dynamic) loading experiments

Mixing of the cement was carried out on a glass slab at room temperature 22+/-1 °C using the manufacturer's recommended powder: liquid ratio. One gram of liquid was calibrated using a Monoject syringe of 3.0 cc capacity and it was found that one gram represented 1.0 cc of liquid. A scoop was used to dispense the powder and this was calibrated by repeating 10 measurements of the weight of powder dispensed by the scoop. The weights in grams were recorded to two decimal places using a balance (PF-150c, Shinko Denish

LTD, 3-9-11 Yushima,Bunkyo-ku,Tokyo 113-0034) and were shown in Table 6.4. The scoop chosen gave a mean weight of 0.22 gram.

Trial no.	1	2	3	4	5	6	7	8	9	10	Mean	St. Dev.
Weight (gr.)	0.22	0.23	0.22	0.22	0.22	0.22	0.22	0.23	0.22	0.22	0.22 (gr.)	0.00423

Table 6.4: Mean weight scoop measurements (grams)

For each cementation, four scoops of powder were mixed with 0.3 cc liquid. The powder was divided into four equal increments and mixed with the liquid over a wide area of the glass slab. Each increment of powder was brought into the liquid at 15 second intervals until all the powder had been incorporated. The time from the addition of first increment until completion of the mix was 1.5 minutes. The cement and cementation armamentarium is shown in Figure 6.12.

Three different techniques were used for cementation:

(1) Un-cemented (UC):

Some experiments were carried out without cement between the gold crown and nickel chromium die.

(2) Partial cementation (PC):

A thin layer of the luting cement was painted onto the axial surface of the die with a fine brush to within about 1mm of the occluso-axial line angle without the occlusal surface being covered. The crown was then seated on the die using axially directed digital pressure.

(3) Full cementation (FC):

The crown was half filled with cement before seating on the die using axially directed digital pressure.

In each case, the orientation of the crown was determined by aligning the thin line scribed on the external surface of the crown with that scribed on the die when the crown was seated.

6.4.1. The cementation jig

Immediately after seating the crown, the crown-die assembly was transferred to the cementation jig and a 5 Kg axial load applied in the middle of the occlusal surface of the crown for 2 minutes. Once the cement had set, any marginal excess was removed with a sharp explorer and the central screw was removed from the die. The crown-die assemblies were stored dry for 24 hours prior to testing on the bench.

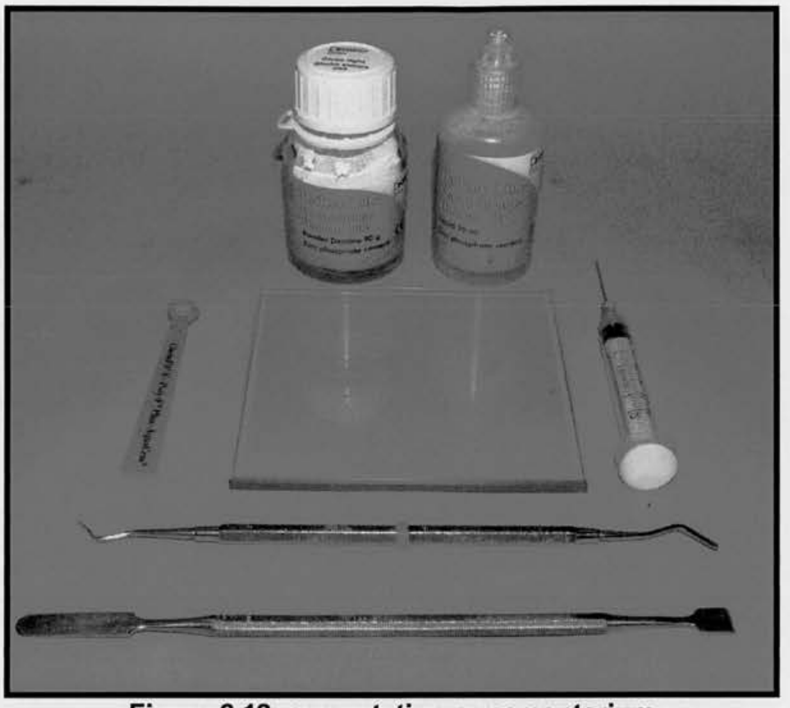


Figure 6.12: cementation armamentarium

6.5. Load application

Axial load was applied at the centre of the occlusal surface of the gold crown using an Instron universal testing machine (Series 5560, Instron Corp., MA,USA) (Figure 6.13). The applied load was measured by a load transducer (load cell) mounted in the moving crosshead of the Instron. The Series 5560 load frame was a structure that consisted of two vertical ball screws, a moving crosshead and a base plate which was its structural foundation.

Some modifications were made to the upper and lower jaws of the Instron machine. For the upper jaw, a small pointed ball-like indenter resembling a tooth cusp was connected (Figure 6.14) to be used for the static loading experiments. For the dynamic loading experiments however a new flat surface indenter of circular shape with diameter of 6.35mm in cross section was fitted and used. Additionally, in the lower jaw, an adaptor was machined that covered the lower jaw base to accommodate a circular housing for the base of the nickel-chromium die.

6.5.1. Instron Marlin Software

Marlin is a software package for materials' testing designed to operate in the Microsoft Windows® environment and allowed capture and analysis of the data. Marlin runs on a computer with a high speed GPIB (HS488). The testing software provided all the test setup, control and analysis functions.

6.5.2. Load application protocol

Depending on the experimental design, the load was divided into two categories:

1. Static.
2. Dynamic.

These were under the control of the software of the Instron machine.

Static loading was controlled by selecting the work task within the compression test option. The magnitude of the load could be varied. At the end of the load application, the return to zero was achieved by selecting the option of stop then return to finish. For all static loading tests, this cycle was repeated five times.

The load magnitude was increased incrementally by 10N up to 220N for most of the static experiments; however, for the experiment examining the effects of heat treatment and increased occlusal thickness of gold crowns to 1.5mm the load was increased by 20N increments up to 460N. The crosshead speed was 0.5mm/min.

Dynamic loading was controlled by selecting the cyclic loading programme. This moved the upper jaw of the machine up and down with the amount of displacement of the movable upper jaw controlled by the distance or load magnitude. For these experiments, the load magnitude was between 0-450N. The frequency was 2Hz.

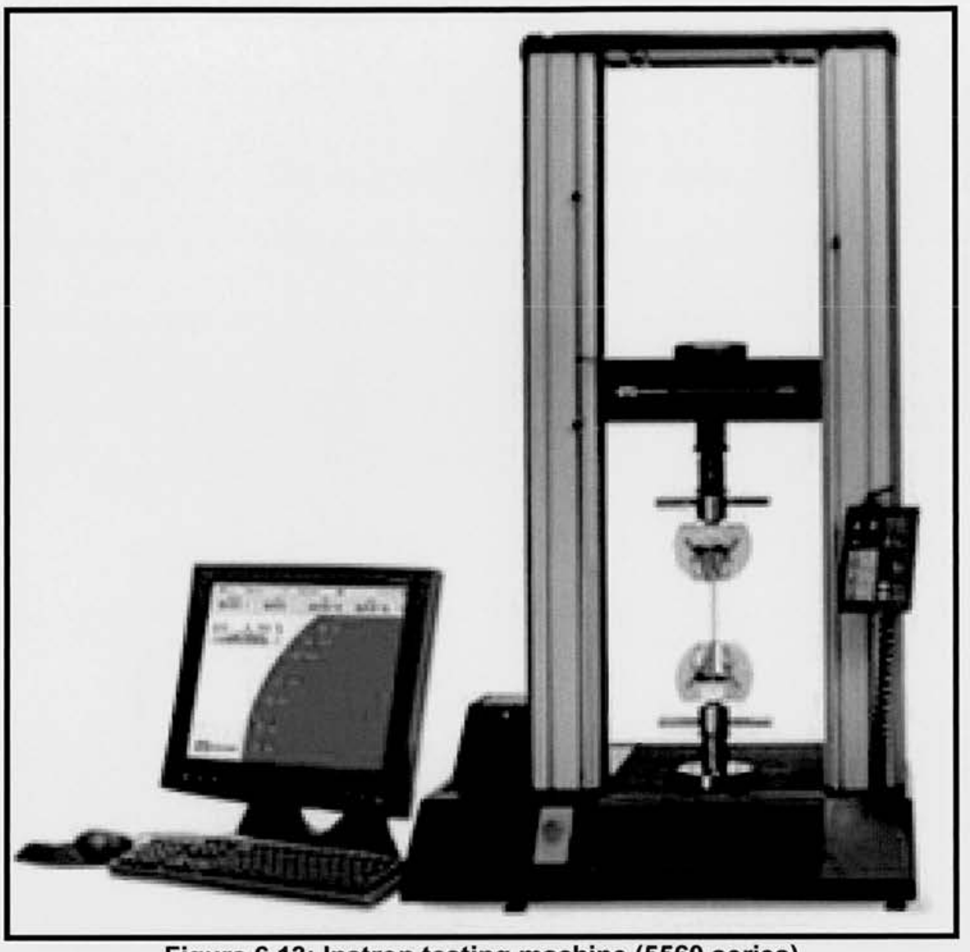


Figure 6.13: Instron testing machine (5560 series)
(www.instron.co.uk/wa/products/universal_materials/5560.aspx)

Figure 6.14: Sample under testing

6.6. Data collection and manipulation

The strain data were collected via the connection of the three wire circuit for each active gauge and displayed through the amplifiers, Figure 6.15.

Data were collected for:

1. The static loading experiments.
2. The dynamic loading experiments.

The data for static load experiments were collected manually, directly from the two amplifiers' screens and recorded in volts to three decimal places. The collected data were entered into Microsoft Office Excel 2003 and then manipulated.

In the later dynamic loading experiments, the data were acquired from the two strain gauges at a sampling rate of 100Hz using a data acquisition card (DAQ) with specifications of NI USB-6008/6009, 8 Inputs, 12-bit, 10 kS/s Multifunction I/O (National Instruments Corporation (UK) Ltd. Measurement House, Newbury Business Park, London Road, Newbury, Berkshire RG14 2PS) (Figure 6.16). The data acquisition card was based on Lab VIEW 7.1 version software.

The applied load was cycled at 2Hz between 0-450N for continuous time periods of up to 72 hours for each sample. Because of this duration, the size of the resulting data files exceeded the limits of the available processing capabilities. Therefore, the files were divided into 6-8 sections, with analysis conducted on each before concatenation to create the overall results.

Analysis was performed using the Matlab mathematical software package (MATLAB, Student Version 7.0.1 software, release 14 with service pack 1).

A custom designed MATLAB program was produced which determined the maximum, minimum and mean values together for each individual loading cycle for each crown. These values were also plotted in volts/1000 values. Maximum, minimum and mean values were calculated and plotted for each sample. The data were calibrated according to the calibration factors as microstrain units (µ).

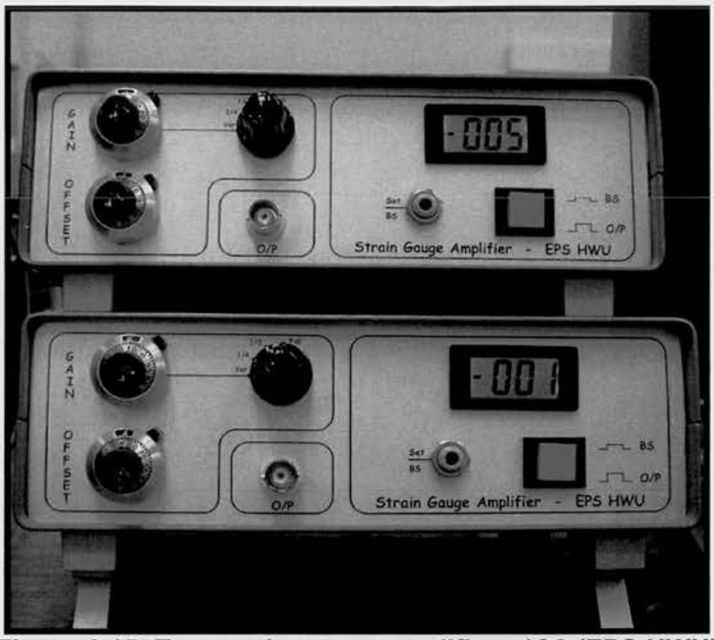


Figure 6.15: Two strain gauge amplifiers 1&2 (EPS HWU)

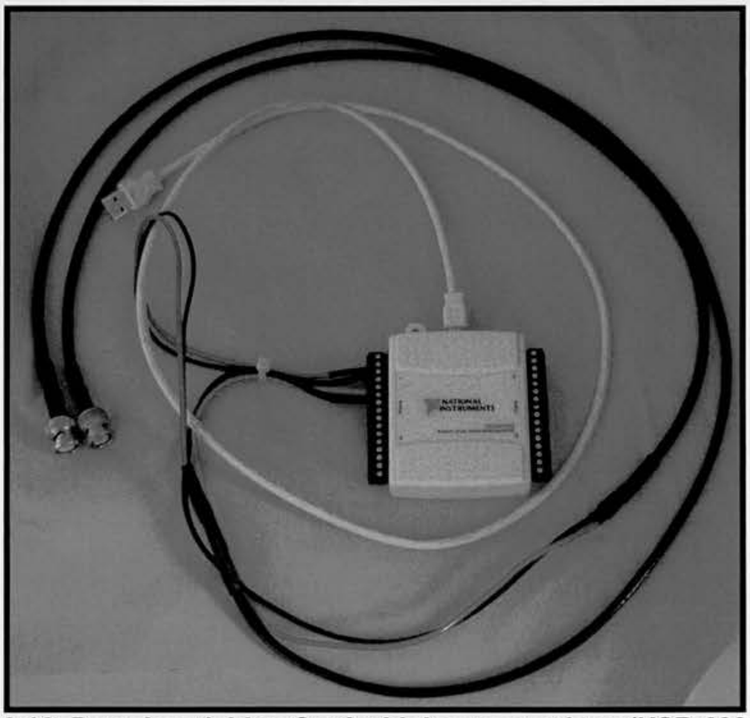


Figure 6.16: Data Acquisition Card with its connections (USB-6008/6009)

6.6.1. Amplifier calibration

The strain gauges amplifiers were calibrated twice, first at the end of the static loading experiments and secondly at the end of the dynamic loading experiments. A metallic aluminium copper alloy strip (alloy 2024 Callister) of known elastic modulus (72.4 GPa) was used and tested in tension and compression (Figure 6.17). A strain gauge was installed and connected to this metallic strip in a similar way that was already described for gold crowns. The metallic strip with its strain gauges was located in the Instron testing machine and the calibration was carried out (Figure 6.18).

Two calibrations were carried out with care taken to ensure that the strain remained below the yield point (approximately 2000 microstrain) during load application.

Measurements were made first under a tensile load which increased incrementally by 50N from 0 to 950N and then decreased incrementally to zero. The loads were recorded manually, plotted in Microsoft Excel tables and used to calculate the calibration factor for each amplifier and plot their calibration curves. The measurements when increasing and decreasing the loads were highly reproducible with little or no hysteresis and were shown in Graph 6.1 and 6.2.

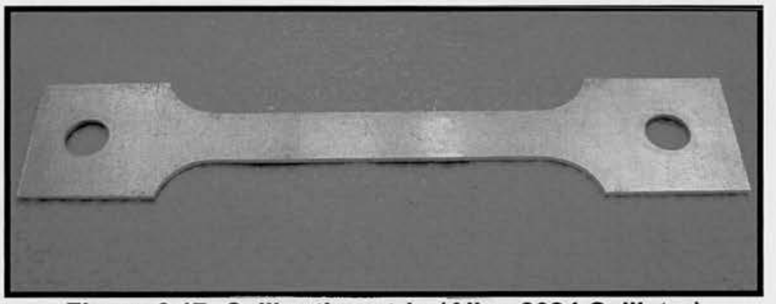
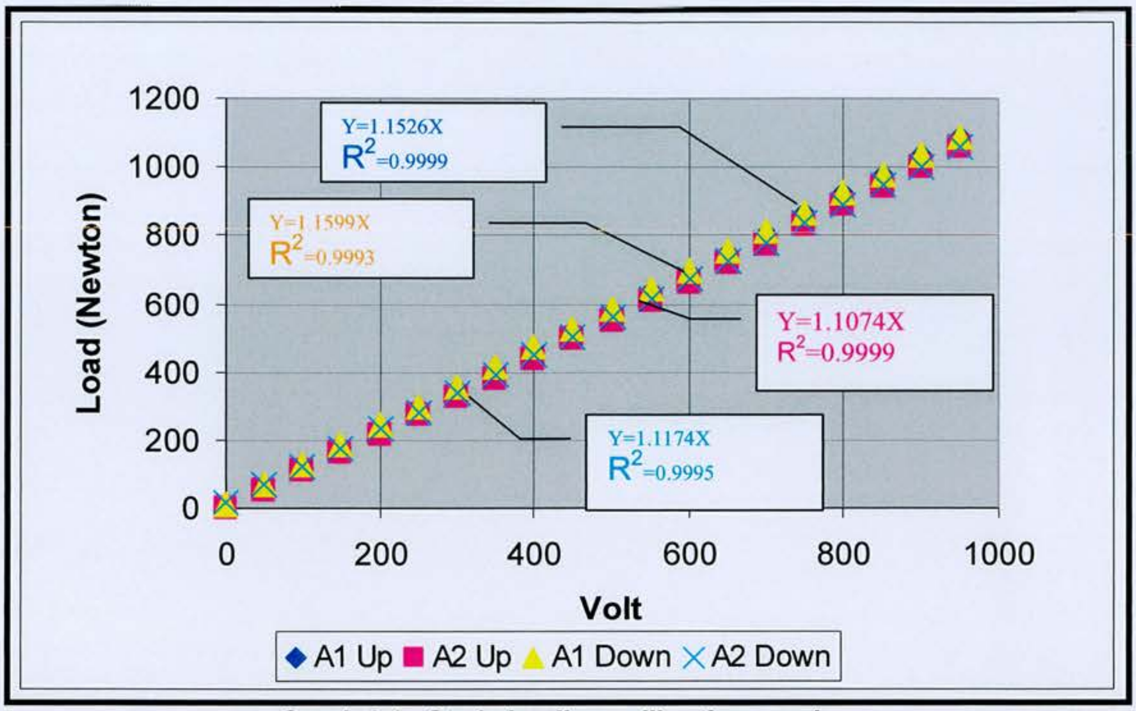
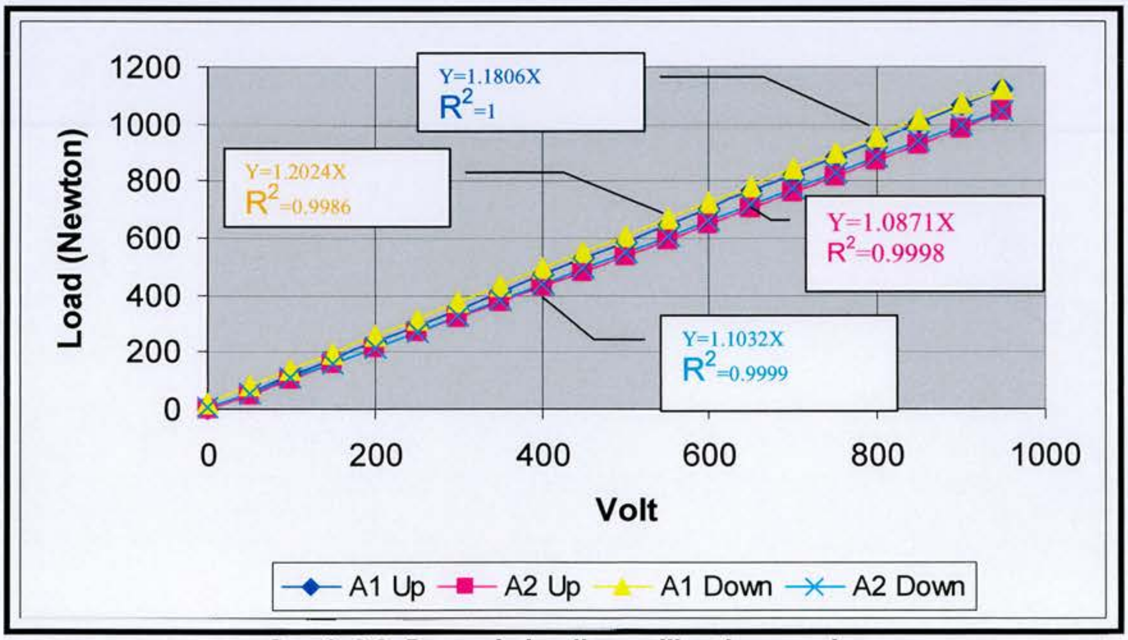


Figure 6.17: Calibration strip (Alloy 2024 Callister)

Figure 6.18: Alloy 2024 Callister with strain gauge installed during amplifier calibration



Graph 6.1: Static loading calibration graph.



Graph 6.2: Dynamic loading calibration graph.

6.7. Crown-die assembly sectioning and polishing

6.7.1. Sectioning

The crown-die assemblies tested under dynamic loading were later examined using a scanning electron microscope. The cyclic loaded samples were embedded in a clear chemically-curing resin material, (Bondaglass-Voss Ltd. 158-160 Ravenscroft Road. Beckenham BR3 4TW).

The samples were placed in a silicone mold (Figure 6.19) which was designed to accommodate two samples at the same time. The resin was mixed according to the manufacturer's instructions and poured into the silicone container so that the pair of samples was completely covered. It was left for 24 hours to cure. The hardened resin was removed from the container, finished, polished (Figure 6.20).

Each finished sample was then placed in a sectioning machine (Isomet 1000) (Figure 6.21) and sectioned longitudinally under lubrication in the middle of each sample through the middle of the strain gauges. The sectioned samples were coated with light cured dental adhesive (Prime and Bond®NT) prior to being polished.

6.7.2. Polishing

Polishing was carried out on an universal polishing machine (METASERVE) using abrasive paper of 600c with water lubrication and then a polishing cloth (PLANO-CLOTH M, Magnetic backed, MetPrep Ltd, Curriers Close, Charter

Avenue, Coventry, CV4 8AW). Final polishing solutions were used which were diamond lubricant and diamond paste of 1 micron particle size. The samples were stored dry within sealed airtight bags.

Figure 6.19: Silicone container

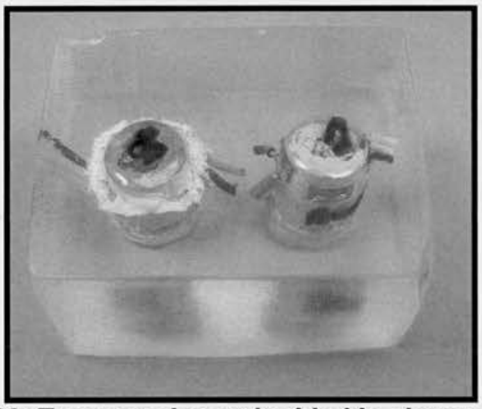


Figure 6.20: Two samples embedded in clear resin block

Figure 6.21: Sectioning machine (Isomet 1000)

6.7.3. Sectioned crown-die assembly scanning electron microscope evaluation

Each sample was examined using a scanning electron microscope (SEM 505 Philips) (Figure 6.22). The sample was first generally scanned to view the entirety of the luting cement layer and then the interface between the crown and the die was viewed at higher magnification of up to 1000-2500 times. Photomicrographs were taken of each sectioned sample. There were stored as digital images.

Figure 6.22: Scanning Electron Microscope (SEM 505 Philips)

Chapter 7: Investigations

7.1. Static loading

7.1.1. Experiments related directly to luting cements

7.1.1.1. Strain behaviour in relation to luting cement distribution

7.1.1.1.1. Cement distribution under cemented cast full crown

Aims:

The aim was to determine the effects of differing cement volumes and methods of application on the distribution of luting cement between the die and the cast full gold crown.

1. To demonstrate and understand the cement distribution between the crown and tooth preparation.
2. To determine a method during cementation in which the axial surfaces of the crown and die but not the occlusal surface were covered by cement.
3. To determine the accuracy and reproducibility of the method.

Material and methods:

One nickel chromium die with its gold crown was used for this experiment. The die had 12^{o} total occlusal convergence (TOC) and 6mm axial height with 11mm diameter as shown in Figure 6.6. The gold crown was of 0.5mm thickness both occlusally and axially.

The crown was located on its die according to the reference lines that were marked on the crown and die.

Cementation technique:

The luting cement was proportioned and mixed, the cement was applied only to the axial surface of the die (Figure 7.1) and the procedure followed those described previously in section 6.4.

The alignment between the crown and the die was verified before placing the crown-die assembly in the cementation rig under a static load of 5kg for 2 minutes. The cemented sample was then moved from the cementation rig and left to set for 30 minutes. Once the cement had set and the excess had been removed, the crown was removed from the die by turning the central screw to push the crown occlusally off the die. The distribution of the cement between the crown and die was assessed visually. The same crown was cemented using the same method on ten occasions.

Results:

The results were illustrated in Figures 7.2-7.11.

In no case did cement cover the entire occlusal surface. However in every case, some cement reached the occlusal surface. In all instances the cement covered the entirety of the axial walls. Six out of ten cementations showed very little spread of the cement onto the occlusal surface.


Figure 7.1: Luting cement painted on the axial surface of the die


Figure 7.2: Trial 1

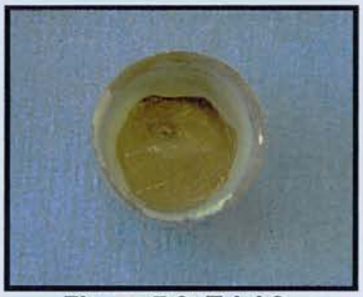


Figure 7.3: Trial 2

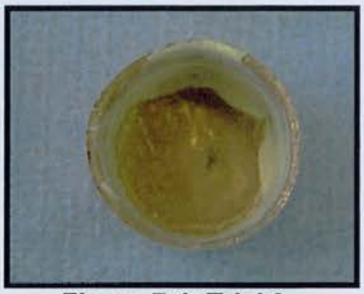


Figure 7.4: Trial 3

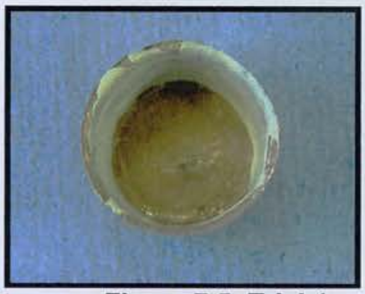


Figure 7.5: Trial 4

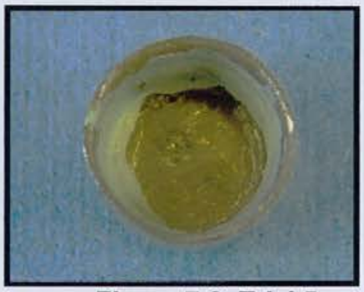


Figure 7.6: Trial 5

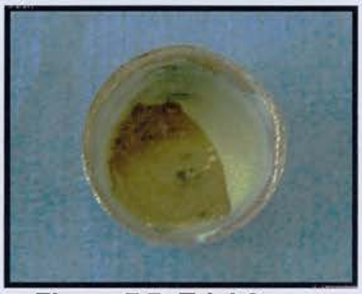


Figure 7.7: Trial 6

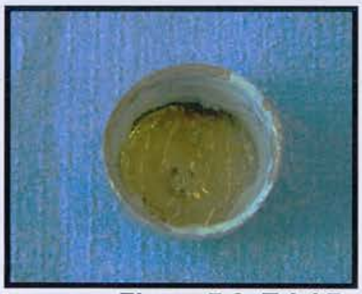


Figure 7.8: Trial 7

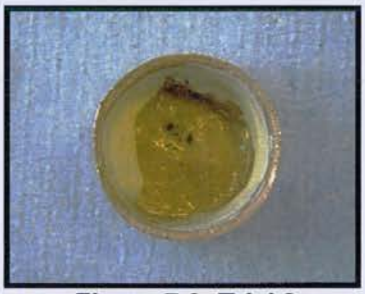


Figure 7.9: Trial 8

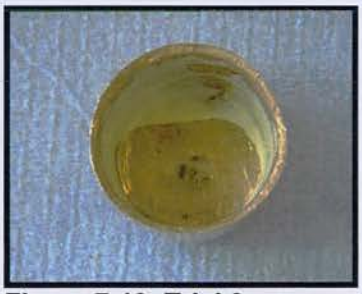


Figure 7.10: Trial 9


Figure 7.11: Trial 10

Discussion:

During cementation of a full cast crown, a pressure is generated as the crown seats over the preparation. This pressure comes from the close proximity of the fitting surface of the crown to the preparation as the crown approximates the finish line of the preparation; in addition the presence of the cementing material with its rheological properties increases such a pressure.

After removal of the cemented gold crown from its die, in all the samples the cement film adhered to the fitting surface of gold crown with only some broken traces on the nickel chromium die.

All the 10 tests showed that it was possible to cement the gold crown on the nickel chromium die without the occlusal surface becoming significantly covered.

It was considered possible that the extent of the cement distribution between the die and the crown might influence the strain in the axial walls of the crown when loaded.

In order to test this hypothesis, it was considered important to be able to control the distribution of the cement film. The results indicated that applying the cement only to the axial walls of the die was effective in minimising its spread on to the occlusal surface.

Conclusions:

The results from this experiment confirmed that applying cement only to the axial walls of the die limited cement spread occlusally on seating the casting with a percentage of 60%. Where occlusal spread was found, this rarely extended more than 1mm across the occlusal surface. It was concluded that this would facilitate the comparative assessment of axial walls strains of non-cemented, partially and fully cemented crowns. This protocol for partial cementation was adopted for use in later experiments.

7.1.1.1.2. Stress-strain behaviour of cast restoration in relation to luting cement distribution

Aims:

1. To determine a method by which the strain developed on the axial surface of the gold crowns cemented on dies could be measured when the crowns were loaded occlusally.
2. To examine the strain on the axial surface of the gold crown resultant from an axial load applied to the occlusal surface of the crown following each cement procedure.

Materials and method:

Four nickel chromium dies with their gold crowns were used. The total occlusal convergence was 12° and axial height of 6mm with 11mm diameter. The overall total thickness of the gold crowns was 0.5mm occlusally and axially. Two miniature strain gauges for each sample were installed on the axial surface opposing each other on the gold crowns as described in Section 6.3. Each crown was tested without cement, partially cemented, and fully cemented on their respected dies as described in Section 6.4.

The nature of load was of incremental type increasing by 10N, from 0N up to 220N. For each cementation, the test was repeated five times for each crown:die assembly. The crosshead speed was 0.5mm/min. The procedure of load application and the machine specification were described in Section 6.5.

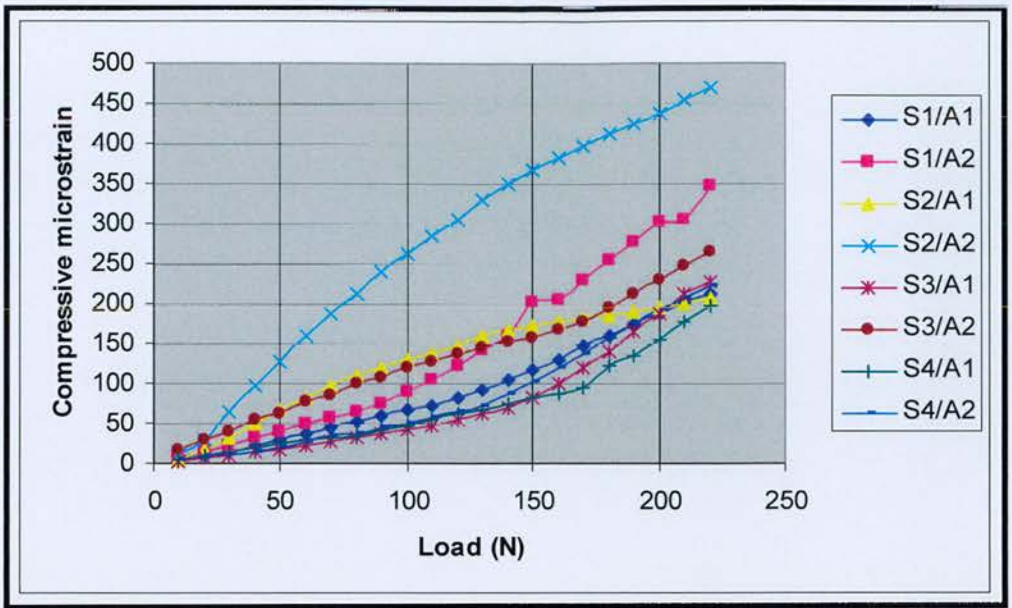
The values for strain were recorded by each amplifier in volts to 3 decimal places. The results for the strain were recorded for each amplifier at each applied load (Newton). All the results were written manually and plotted using an Excel programme. The means were taken for the two strain gauges for each sample. The figures were determined according to the calibration factors of the two amplifiers used in the experiments for static loading.

Results:

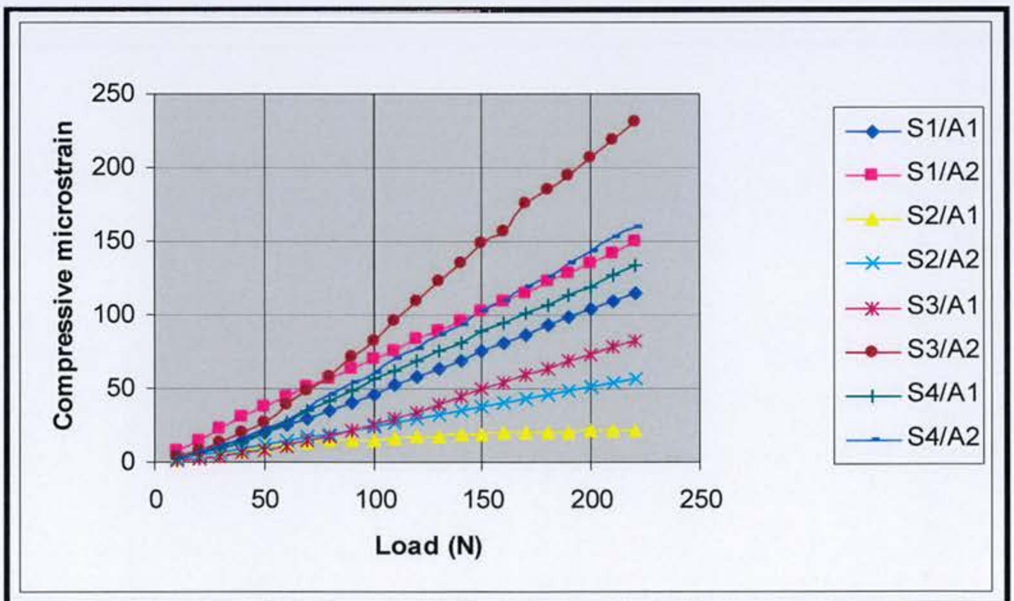
The results were divided into three groups depending on the cementation procedure. These were displayed as graphs of load in Newtons against compressive microstrain (μ).

Graph 7.1 showed the data for the un-cemented samples. Graph 7.2 showed the data for the partially cemented samples. Graph 7.3 showed the data for the fully cemented samples.

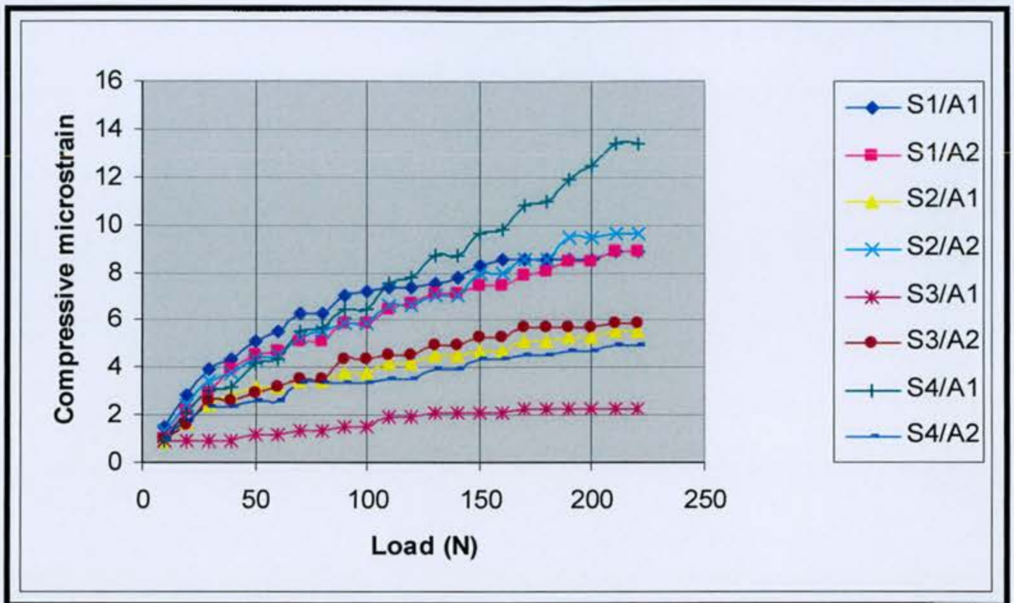
Graph 7.4 showed the mean values for strain for each cementation type plotted against load (Newton). From the data used the linear equation was used to calculate the slope which was displayed as straight lines for each cementation type together with their values.



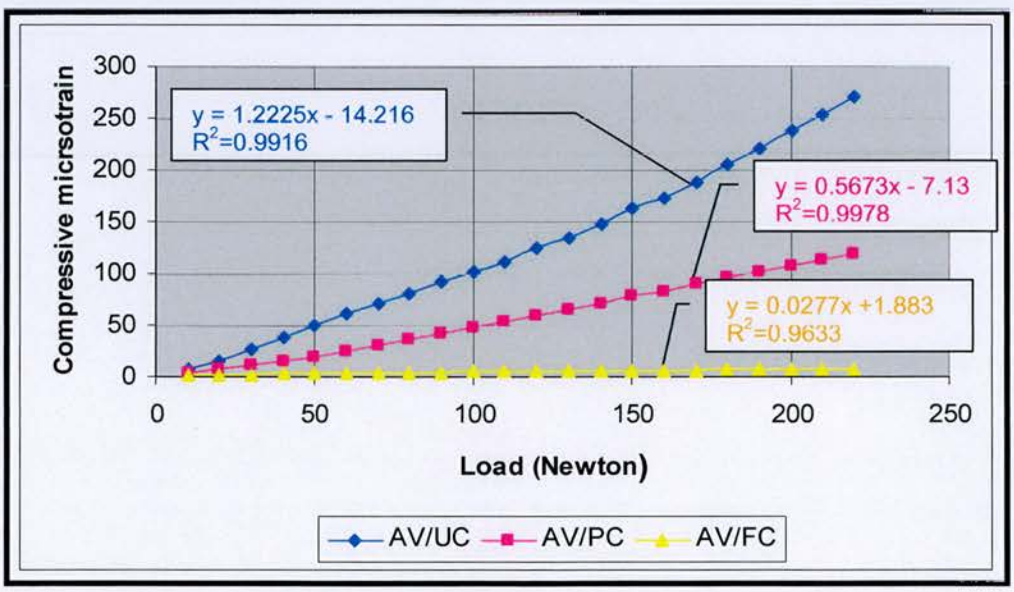
Graph 7.1: Load v compressive microstrain of un-cemented crowns (12° &6mm)



Graph 7.2: Load v compressive microstrain of partially cemented crowns (12° &6mm)



Graph 7.3: Load v compressive strain of fully cemented crowns (12° &6mm)



Graph 7.4: Load v mean compressive microstrain with different type of cementation (12°&6mm)

Discussion:

The mean calibrated values were recorded in Appendix Section 12.1.1. Increasing compressive strain was recorded by the strain gauges as increasingly negative values. For ease of illustration, these values have been shown as positives.

The greatest values for strain were found in the uncemented group. It was of interest that some of the values in the partially cemented group resembled those found in the uncemented group. The reasons for this might have been firstly that there was less cement between the crown and the die than might have been expected or that the absence of cement on the occlusal surface produced a profound influence on the strain in the axial walls of the crown.

The results generally showed a linear increase in the strain as the applied load increased. Within each cementation condition there was also a linear relationship between the applied axial load and the peripheral strain on the axial surfaces of the gold crowns. As the applied load increased, the strain increased. Comparatively, the magnitude of the strain on the axial surfaces of the gold crowns was much higher for the un-cemented crowns compared with the fully cemented crowns. The partially cemented crowns were intermediate. Generally, the load-strain relationship was linear and repeatable and the slope varied over a factor of about 40 between the un-cemented and fully cemented crown and about 2 between un-cemented and partially cemented, demonstrating that there was a clear sensitivity to bond integrity. The equations for all the cementation scenarios were shown in Graph 7.4.

For standardisation purposes, for each of the four dies, the three crowns were cemented using the three different cementation protocols. During cementation of the partially cemented group, the luting agent was applied only to the axial surface of the dies leaving the occlusal surface clear of cement. This process was not absolutely precise in the preventing cement spread completely on to the occlusal surface as described in Section 7.1.1.1.1. This might have explained the relatively large range in the strain results from one die to another. Another factor might have been variations in the adaptation of the crown on the die or skew of the crown during seating. However, the ranges were greater in the uncemented and partially cemented crowns compared with the fully cemented specimens. If skew had been responsible for the wider ranges, this would have been more likely to have been evident in the fully cemented group.

The behaviour of the luting cement appeared to be affected by the cement distribution. It seemed that there was a mutual effect as the presence of the luting cement on the axial surface possibly maintained the integrity of the cement layer on the occlusal surface where there was an axial compressive load. The converse was also probable as the presence of the luting cement on the occlusal surface was vital for the integrity of the cement layer on the axial surfaces. Additionally, the presence of the luting cement on the occlusal surface seemed to be important to decrease the resultant strain on the axial surfaces of cemented crowns. This was demonstrated from the slope of the graphs where the fully cemented varied by a factor of about 0.05 compared with the partially cemented. The combined effect of the presence of the luting cement on both occlusal and axial surfaces resulted in the low strain values in the fully cemented samples.

In the partially cemented samples, the strain was bigger than the fully cemented counterparts. In fully cemented samples, the luting cement on the occlusal surface would have been under compression and on the axial surfaces under tension. This observation showed the importance of the luting cement on the occlusal surface in decreasing the strain on the axial surfaces of cemented crowns. However, in the partially cemented samples, the luting cement was under tension only. This would have explained the significant decrease in the strain magnitude in the fully cemented compared with the partially cemented samples.

The strain on the axial surface of the crowns was influenced by the presence of the luting cement underneath. It confirmed the absorbing effect of the cement layer of the resultant strain produced by the applied load as luting cements are more durable in compression than tension.

The cement lute decreased the axial strain in the gold crown. The effect of cement distribution occlusally on the strain distribution was also clear. The fully cemented crown markedly decreased the strain on the axial walls compared with the un-cemented, whilst the effect was less for the partially cemented crowns.

The crowns used in this series of experiments had been made on spaced dies but without measuring the thickness of the die spacer. It was considered important to control the space that was produced by application of the spacer.

Comparing with other studies in relation to the effect of cement distribution on the occlusal surface and the strain distribution, This observation was in agreement with Van Roekel (1995) who reported that the compressive strength of dental cements was greater that either the tensile or shear strength. Additionally, this piece of work supported Yamashita *et al* (1998) who reported that the existence of luting cement in fully cemented crowns between the restoration and the die significantly affected the strain distribution under load. However, they used a cantilever bridge in their experimental model, which was completely different from the present study where axial loads were applied centrally on the occlusal surface of the crowns.

Conclusions:

With the limitations of this study the following conclusions were drawn:

1. The strain gauge technique was a reproducible method for determining the strain on the axial surfaces of crowns.
2. There was a general linear increase in the axial strain with increase of the applied axial load.
3. It was evident that the strain on the axial surfaces of the gold crown decreased with the increase in the extent of the cement film between the crown and die.
4. The presence of luting cement on the occlusal surface had a major effect on the strain distribution of the cemented crowns.
5. The luting cement acted to reduce the resultant strain produced by the applied load between the crowns and their nickel-chromium dies.

7.1.2. Strain behaviour in relation to luting cement thickness:

7.1.2.1. Die-spacer film thickness:

Introduction:

The initial experiments in the static experiments used spaced crowns, but the amount of spacing was not controlled. It was considered important that a known cement space was created between the die and the crown and its effect on the strain distribution.

Aims:

1. To determine the film thickness with the number of applied layers of die spaces.
2. To determine whether there was any difference in the die spacer thickness applied on die stone and cast nickel-chromium alloy.
3. To use the results to standardise the application of die spacer in further experiments.

Materials and method:

The steel shaft of a micrometer screw gauge (Shardlow Micrometers LTD, Sheffield, England) was used to create a master model. The micrometer shank was circular in cross section with diameter of 6.78mm. A cylindrical shape plastic container was used as an impression tray and an impression was made of the stainless steel shank using vinyl polysiloxane impression material (Coltene/Whaledent AG. Feldwiesnstrasse 20, 9450 Altstatten/ Switzerland). A single mix technique was used for the impression where light bodied impression material was used to coat the micrometer shank whilst heavy bodied impression material was loaded into the tray. The shank of the micrometer was pushed into the impression material inside the impression tray to a depth of approximately 10cm. The impression was left to set for 20 minutes. The shank and the impression were removed as one unit from the impression tray, and then the shank was separated from the impression. The impression was sectioned longitudinally using a scalpel blade but the base was left uncut. The impression was left for one day before being returned to the tray and poured in type II die stone (Whip Mix®, Whip Mix Corporation. P.O.Box 17183. Louisville, KY 40217-.0183 USA).

The powder to liquid ratio followed the manufacturer's instructions of 25 gram of powder mixed with 10cc of water. The mixing was under vacuum (Refer, Twister Pro, Germany) for one minute, before pouring under vibration. The stone was left to set for an hour then separated from the impression and left to finally harden for 24 hours.

The diameter of the stone model and steel shank were measured using an electronic digital calliper. The points of measurements were controlled by reference to the cross mark drawn on the top of the stainless steel shank and the stone rods.

Measurements were recorded in millimetres to two decimal places. The ten stone rods had a mean diameter of 6.81mm. Figure 7.12 showed the steel shaft and its stone replica, both coated with silver coloured die spacer.

Tru-fit silver coloured die spacer was used in the experiment (TRU-FIT, Geo. Taub products& fusion co. inc. Jersey City, N.J., USA), (Figure 7.13) together with its proprietary thinner. Before use, the die spacer was shaken for 30 seconds. The die spacer was dispensed and mixed with the thinner in a dappen's dish at a ratio of one dip with the brush into the spacer and two dips into the thinner.

The die spacer was painted onto the circumference of the stone rod and the stainless steel shaft in a single stroking motion. Each layer was allowed to dry before applying the next. The caps on the bottles were immediately replaced after each application, and the brush cleaned frequently with the thinner in a separate dish. The thickness of the stone rod was measured 5 minutes after the application of each layer of die spacer painted using the same protocol as for the initial measurements. Eight layers of die spacer were applied to the stone rods and the stainless steel shaft.

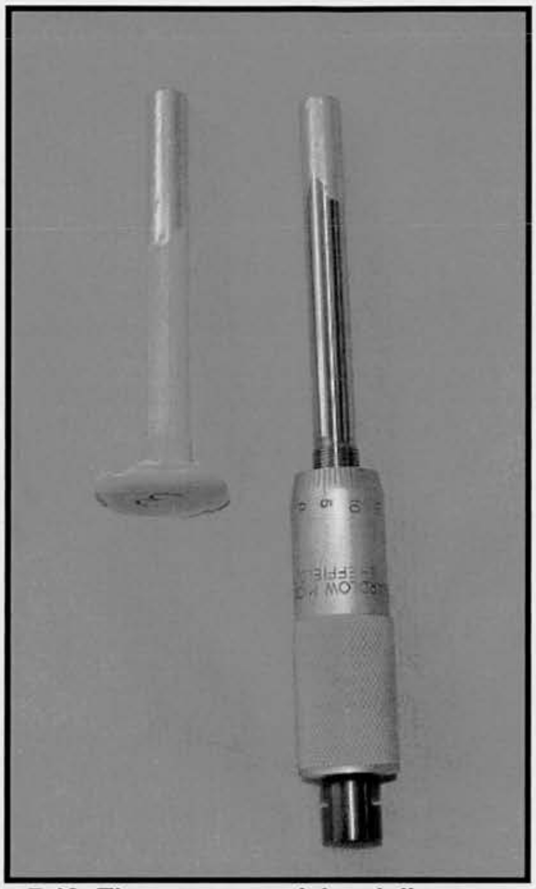


Figure 7.12: The master model and die stone model

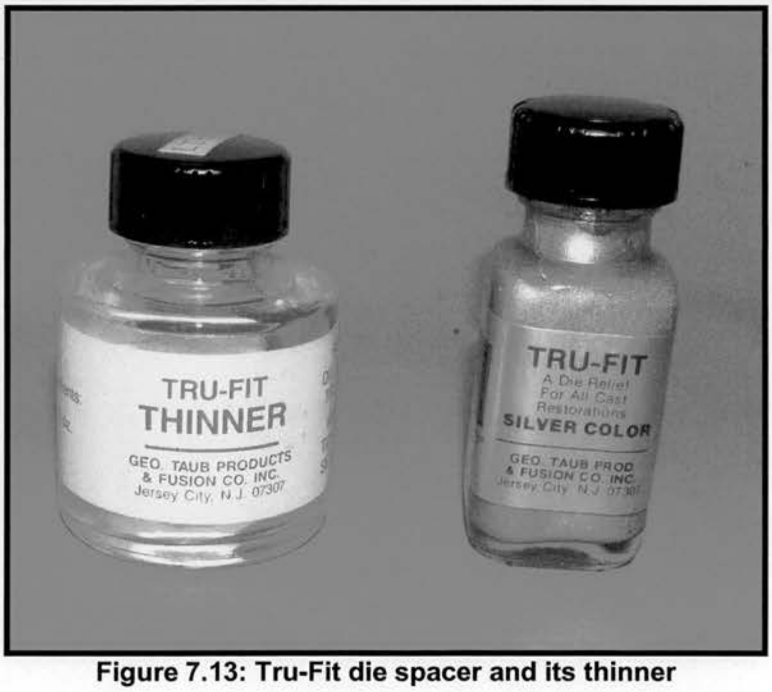


Figure 7.13: Tru-Fit die spacer and its thinner

Results:

The results for the thickness of each stone sample were recorded in micrometres and shown in Table 7.1.

The thickness for each die-spacer layer for each on stainless steel shank was determined and shown in Table 7.2.

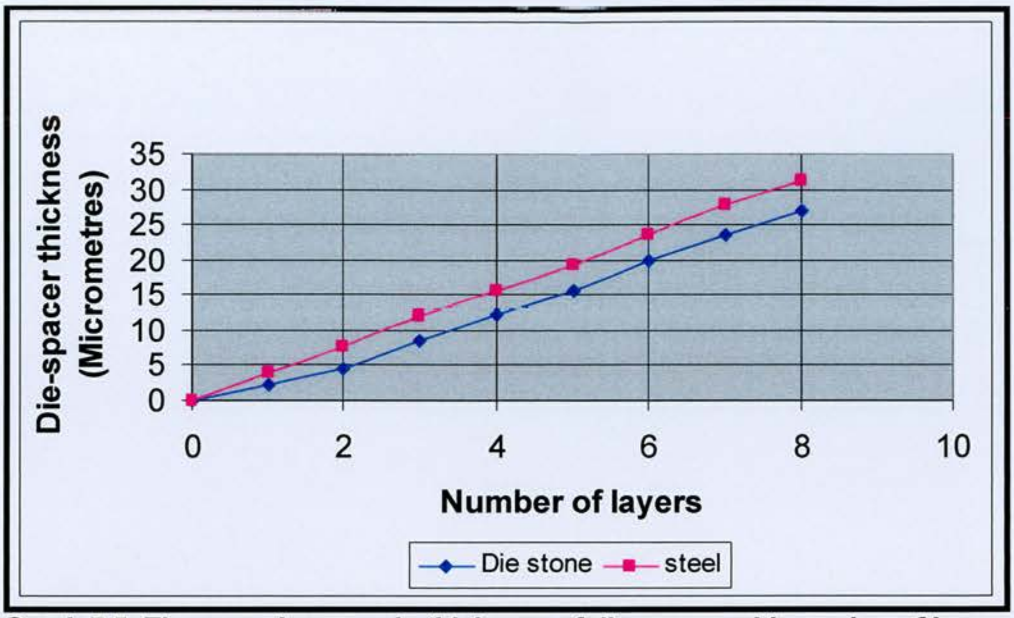
The mean thickness increase of die spacer with increased number layers application for both stone dies and steel rod was shown in Graph 7.5.

		Samples										Mean	S.D.
		1	2	3	4	5	6	7	8	9	10		
Layers	0	0	0	0	0	0	0	0	0	0	0	0	0
	1	3	2	3	2	2	2	2	2	2	2	2.2	0.423
	2	5	4	5	5	4	6	4	4	4	4	4.5	0.707
	3	9	7	10	9	9	9	7	8	9	9	8.6	0.966
	4	12	12	12	13	13	13	11	12	12	12	12.2	0.632
	5	14	16	14	16	18	17	14	18	14	16	15.7	1.636
	6	18	19	18	23	22	23	18	21	18	19	19.9	2.132
	7	21	22	23	28	26	27	21	24	23	21	23.6	2.591
	8	24	24	27	32	31	32	23	28	25	24	27	3.559

Table 7.1: The die spacer film thickness (Micrometer) with number of layers on die stone

		No. of application										Mean	S.D.
		1	2	3	4	5	6	7	8	9	10		
Layers	0	0	0	0	0	0	0	0	0	0	0	0	0
	1	2	3	3	3	5	5	5	5	5	5	4.1	1.197
	2	6	7	5	6	8	8	9	9	10	9	7.7	1.636
	3	10	10	10	10	12	13	14	12	14	14	11.9	1.792
	4	14	14	15	14	17	16	16	17	17	17	15.7	1.338
	5	20	17	18	17	23	18	19	20	20	21	19.3	1.889
	6	24	21	23	22	27	24	23	24	25	24	23.7	1.636
	7	28	26	28	25	32	29	27	27	29	28	27.9	1.912
	8	30	29	29	27	37	35	31	31	32	31	31.2	2.936

Table 7.2: The die spacer film thickness (Micrometer) with number of layers on stainless steel shank



Graph 7.5: The mean increase in thickness of die spacer with number of layers applied for die stone and steel rod.

Discussion:

The results showed that there was a broadly linear increase in the diameter of the stone rods and the stainless steel shank with the increased number of layers of die spacer applied. However the increase was greater on the stainless steel shaft than on the stone rods. With eight layers of die spacer on die stone there was increase of 27µm, whilst on stainless steel it was 31µm. The mean increase in the thickness on the stainless steel averaged nearly 4 µm for each layer of die spacer, whilst for the stone rods this was just 3.4 µm. The differences might have been related to the ability of the stone to absorb the die spacer.

There was a clear trend showing that application of successive layers of die-spacer produced small and relatively consistent increases in thickness.

However, the nature of the measuring technique produced limitations. The ability of a dial gauge to record accurately measurements in single figure micrometres was highly questionable. However, the standard deviations were generally of a low order particularly at greater film thickness.

The data enabled the adoption of a standardised application technique for use in this study.

Conclusions:

There was an increase in the die spacer thickness with the increase of the number of layers applied. This was broadly linear demonstrating that well-thinned die spacer produced a generally consistent thickness in successive layers. This protocol for application was adopted in the construction of the crowns used in this study.

7.1.2.2. The strain-strain behaviour of cemented gold crown made with different degrees of spacing

Aims:

The aim of this piece of work was to investigate the effect of die-spacing on the strain response of the cemented crowns.

Materials and method:

The experimental samples were five nickel chromium dies with total occlusal convergence of 12° and 6mm axial wall height. Fifteen cast gold crowns were fabricated. The first group was fabricated without die-spacer application, the second group with 10 layers of die-spacer, and the third group with 20 layers of die spacer.

The method started with fabrication of five gold crowns (one on each die) without die-spacer applied on the nickel chromium dies.

For the other crowns with die-spacer application, the screw was fitted into the tapped hole of each nickel chromium die to make it flush with the occlusal surface to simulate a flat occlusal surface as described in Section 6.2. After thorough mixing, the die spacer was painted on the dies circumferentially, to cover the entire axial and occlusal surfaces. After each layer, the die-spacer was given two minutes for drying, layer by layer up to the required number of layers.

For the relieved dies, a no. 10 scalpel blade was then used to delineate the die spacer 1mm from the preparation's finish line.

Three gold crowns were waxed for each of the five dies in each of the three groups. The procedure of gold crown fabrication was described in Section 6.2. A total of fifteen gold crowns were produced. The produced gold crowns had an overall thickness of 0.5mm.

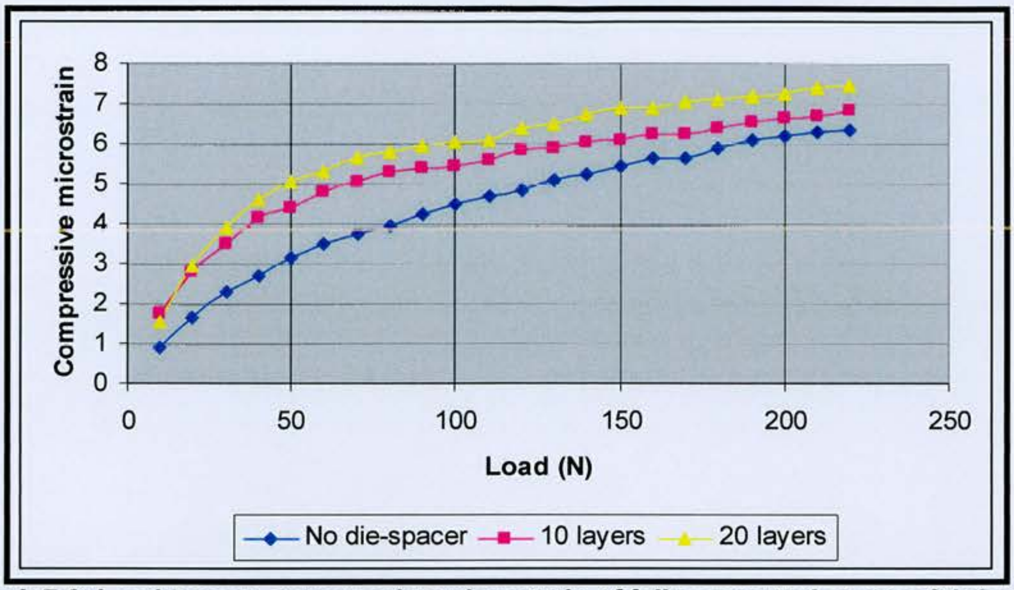
Two strain gauges were installed on each gold crown approximately 1mm above the crown margins opposing each others with their two solder terminals. This was described in Section 6.3.

All the samples were fully cemented with each crown being half-filled with cement before seating following the technique described in Section 6.4.

Using the Instron Testing Machine, a compressive load was applied in increasing 10N increments from 10N upto 220N. The cross-head speed was 0.05mm/min. The loading was repeated five times for each sample. The strain was recorded following the method described in Section 6.5.

Results:

The results for each group of cemented crowns were displayed in Graph 7.6 showing the mean compressive microstrain as positive values plotted against load in Newtons.



Graph 7.6: Load v mean compressive microstrain of fully cemented crowns fabricated with different layers of die-spacer.

Discussion:

Each of the five dies received three cemented crowns in turn; the first five crowns were un-spaced, the second relieved by 10 layers of die spacer and the third relieved by 20 layers. According to the calibration study described previously (Section 7.1.2.1) 10 layers die-spacer represented about 40µm thickness and 20 layers die-spacer about 80µm.

The mean was calculated for the five runs for each sample in their category (no die spacer, 10 layers die spacer, and 20 layers die spacer). The results generally showed that with the increase of the applied compressive load, there was an increase in the resultant strain on the axial surface of the gold crowns for the different degree of spacing.

The use of no die spacer resulted in the lowest strains. There was a small increase in the strain with the use of die spacer. This was more evident for 20 coats of spacer than 10. However the differences between 10 and 20 of spacer were generally small whilst the greatest difference was seen between un-spaced and the 10 coats of space crowns. These differences were more marked at lower loads.

The die spacer film thickness was known for the stone rod but the thickness of the cement film beneath each crown was not because of the hydrostatic forces produced on seating. The luting cement film thickness in this study was based on the space that is provided by the number of layers of die spacer. One reason for the strain measurements being different between the 0 layer die-spacer, 10 layers die-space and 20 layers die-spacer is that the cement film thickness between the crown and the die was different. However, as previously discussed the differences between the two spaced groups were small. This might have been because the difference in the film thickness of the applied die spacer in the two groups was less than might have been expected. Alternatively, it would have been anticipated that the presence of die spacer would have improved the seating of the crowns on cementation as compared with the unspaced group. Improvements in seating potentially would have reduced the cement film thickness; however, there was an evident difference between the behaviour of the spaced groups compared with the unspaced. This was probably indicative of an increased cement film thickness in the two spaced groups. Thicker films of cement appeared to be associated with increased axial strain in the walls of the crown.

It was considered that variations in the cement film thickness might influence the axial strain recorded in the crown. It was therefore appropriate to minimise variability in the thickness of the cement beneath each crown.

There was no need to know the actual cement film thickness beneath the crowns as long as the die-spaced layers were consistent. The protocol was established from the study of the measurement of the die spacer on the cylindrical rods (Section 7.1.2.1) which indicated that generally consistent results could be obtained. However, the thickness of the cement film would also have been influenced by the variables associated with the cementation of crowns and would have been likely to have been greater than the relief provided by the die spacer.

Methods have been described for measuring the actual thickness of a film beneath crowns some involve retrieval of a silicone film or sectioning of the crown to allow direct visualisation. Neither was considered suitable for use in this study.

There are contradictory data relating to luting cement film thickness and retention of crowns as a function of die spacer, Eames *et al* (1978), Marker *et al* (1987), Carter and Wilson (1996) found an increase in retention of cemented crowns as the film thickness increased. Contrastingly, Vermilyea *et al* (1983) and Gegauff and Rosenstiel (1989) reported a decrease in retention. At the same time, Hembree and Cooper (1979) and Passon *et al* (1992) did not report any changes. Carter and Wilson (1996) related such variations in the results to variations in the experimental designs. The data from the present study recorded an increase in axial compressive microstrain which would be undesirable.

Conclusions:

With the limitation of this study, the following conclusions could be drawn:

1. Strain measurements from the axial surface of gold crowns could be used to assess the effect of increased cement film thickness.
2. Strain gauges as a measurement device was accurate enough to assess the small differences of cement lute thickness created by measurable layers of die spacer.
3. Increased thickness of cement was associated with increased axial wall strain particularly at lower loads.

7.1.3. Experiments related to preparation geometry

7.1.3.1. Occlusal convergence and strain distribution on axial surface of cemented gold crown

Aims:

1. To determine and evaluate the resultant strain in opposing axial walls of a cemented full gold crown with the applied load using a die of 24° total occlusal convergence (TOC) and 6mm axial wall height.
2. To compare the results of the load-strain response of 12° total occlusal convergence (TOC) with 6mm axial wall height.

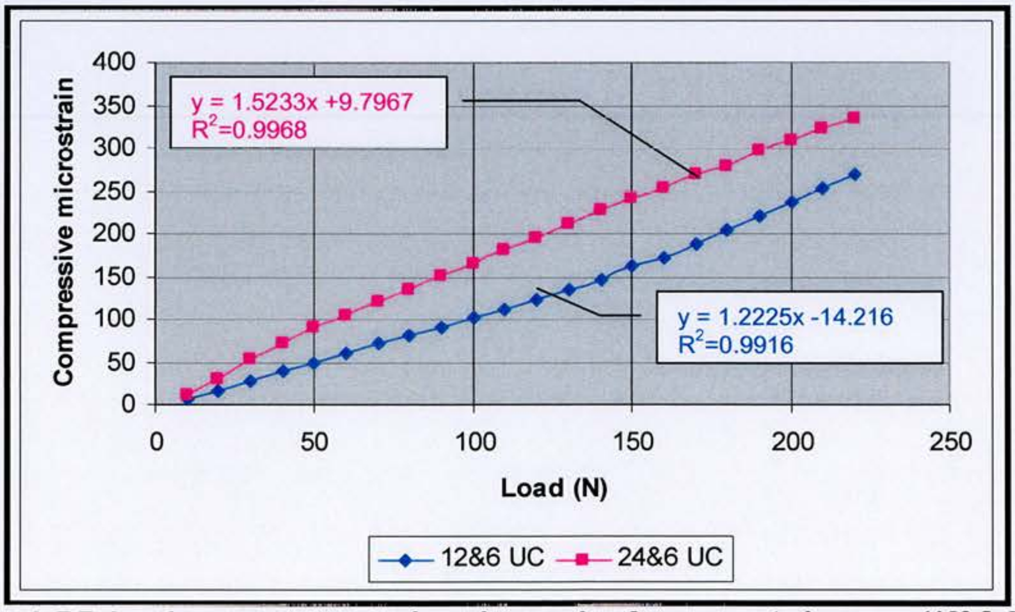
Materials and method:

Five nickel chromium dies with 24° TOC and 6mm axial wall height were produced using the method described in Section 6.1. On each spaced die, a gold crown with 0.5mm thickness was fabricated as described in Section 6.2. Two strain gauges were installed on the axial surface of each gold crown opposing each other as described in Section 6.3. The experiment was carried out using the same protocol of crowns without cement, partially cemented and fully cemented. Each of the five crowns was tested on three occasions once uncemented, once partially cemented and once fully cemented. The cementation procedure was described in Section 6.4.

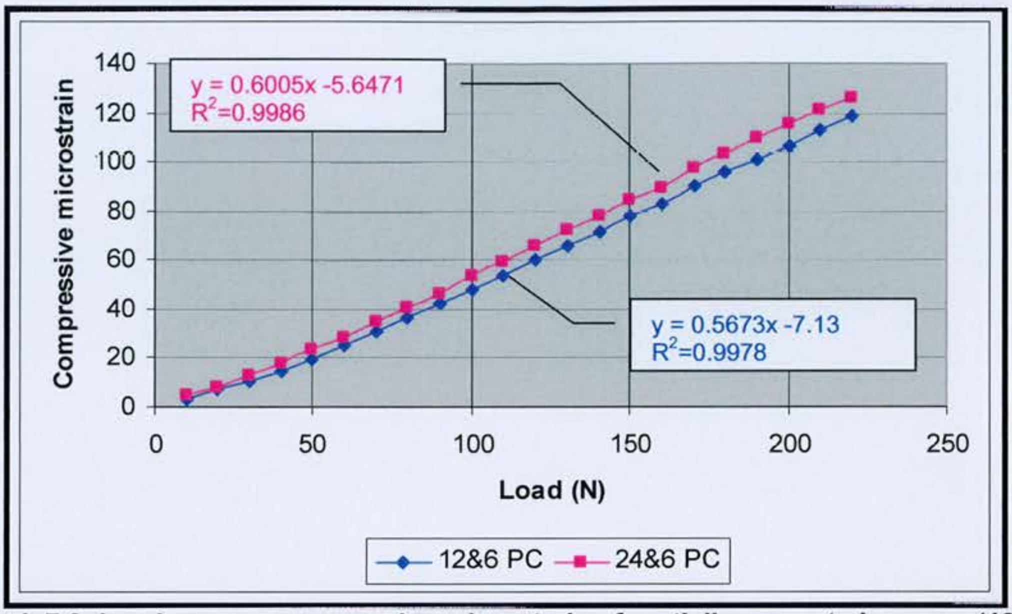
Crowns were tested under compressive axial loads as described in Section 6.5. The compressive load was applied using incremental loading up to 220N using the same method described in Section 6.5. The resultant compressive strain was recorded as described in Section 6.6.

Results:

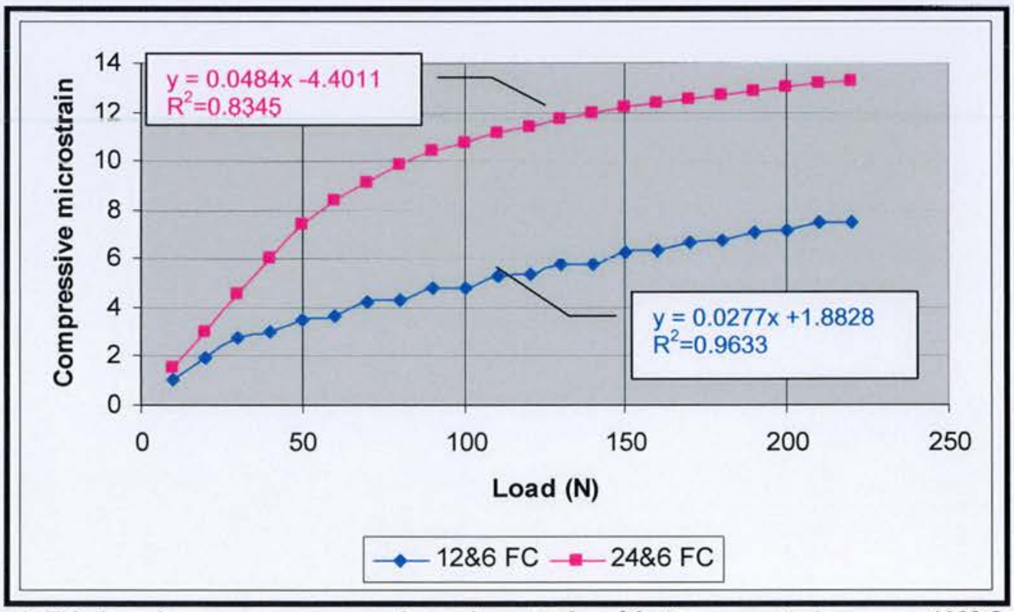
The results for the 12° and 24° TOC were displayed in Graph 7.7 for the uncemented crowns , Graph 7.8 for the partially cemented crowns and Graph 7.9 for the fully cemented crowns. The results were plotted as load in Newtons against mean compressive microstrain shown as positive values.



Graph 7.7: Load v mean compressive microstrain of un-cemented crowns (12° & 24° TOC).



Graph 7.8: Load v mean compressive microstrain of partially cemented crowns (12° & 24° TOC).



Graph 7.9: Load v mean compressive microstrain of fully cemented crowns (12° & 24° TOC).

Discussion:

Two convergence angles were investigated 12° and 24°. The first was larger than the ideal but was still more towards the lower end of that commonly produced by clinicians. The second taper was probably more clinically representative. The results for the two tapers were different, increased taper was associated with increased strain on the axial surface of the crowns.

During preparation of the master stone die preparations, great care was taken to maintain the same height and base diameter whilst increasing the taper from 12° to 24° TOC. However, it was recognised that, as a result of the increase in TOC, the diameter of the die reduced towards the occlusal surface and this also reduced the surface area of the preparation.

The results showed that the strains on the axial surface of the uncemented, partially cemented, and fully cemented crowns with 24° TOC were greater than for the12° TOC crowns.

In graph 7.7 (note the scale for Y axis), the uncemented crowns, the slope of the linear equation for 12° TOC was higher than that of 24° TOC by factor of 1.25. Some of this was probably due to:

1. The diameter of 24° taper crown at the occlusal surface.
2. 12° taper presents walls that are closer to the direction of the applied load and therefore better able to resist strain than 24° TOC.

The responses were broadly linear. Maximum of the mean-values for microstrain was approximately 260 for 12° taper and 330 for 24° taper.

In graph 7.8 (note the scale for Y axis), the values of load against mean of the microstrain for the partially cemented crowns on the dies with 12° and 24° TOC were generally linear and similar, in addition, their slopes were similar. The maximum mean-value for microstrain for the crown with 12° taper was 120 and for 24° taper approximately 130.

Graph 7.9 (note the scale for Y axis) showed that the mean values for microstrain in the fully cemented samples on both dies of 12° and 24° were extremely low. A mean maximum of 8 microstrain was recorded for 12° and approximately 13 micorstrain for 24°.

The characteristic of these graphs were different. For 12° taper, the small increase in strain between 0 and 220 Newtons was generally linear whereas for 24° taper, it was not. In the 24° taper group, loading between 0 and 100 Newtons produced more strain than for the 12° group. However, the values of strain seen in the fully cemented group were small (1-7.5 Microstrain).

Comparatively, the taper of the preparation influenced the strain in the axial walls of the crown with increased taper being associated with greater strains. The presence of cement reduced these strains. The observed major difference between cement and no cement was perhaps unexpected.

However, the large difference between partial and full cementation, where strain in the latter group were greatly reduced, was more surprising.
The reasons for this might have been related to the decrease of the surface area of the more tapered crowns and also their less favourable geometry with respect to the direction and point of application of the applied load.

The graphs showed that the relationship of load to strain was almost linear for all cementation scenarios. The strain magnitude for the uncemented crown was greater than the fully cemented with the partially cemented crowns intermediate.

Comparatively, doubling the TOC did not increase the strain proportionally. The slope of the graph for the fully cemented crown of 12° TOC was greater than that for 24° TOC by a factor of 1.75. However, the strain response from the partially cemented crowns resembled the uncemented specimens more closely that the fully cemented crowns. The cementation condition affected axial wall strain more than the taper of the preparation.

Compared with other studies, the results showed that the greater TOC, the more axial strain, q1which might have an adverse effect on the long term survival of a cemented crown in the clinical situation. This study supported Hegdahl and Silness (1977) who concluded that the use of large convergence angles resulted in small resisting areas and recommended avoidance of such high degrees of convergence.

Conclusions:

1. The cementation condition had a large effect on axial wall strain.
2. The total occlusal convergence (TOC) had a small effect on the strain distribution in the axial walls of the crown.
3. The larger the total convergence angle, the higher the strain in the axial walls.
4. The presence of cement reduced the strain in the axial walls of the crowns. The cement minimised the differences in strain between the crowns with 12° and 24° of taper.
5. The differences in strain between the partially cemented and fully cemented were large.

7.1.3.2. Effect of increasing axial wall height of the preparation on strain distribution in the cemented cast gold crown

Aims:

1. To determine the strain in opposing axial walls of a cemented full gold crown under applied load using a die of 24° TOC and 8mm axial wall height.
2. To compare the results of the load-strain response of 24° TOC and 6mm axial wall height.

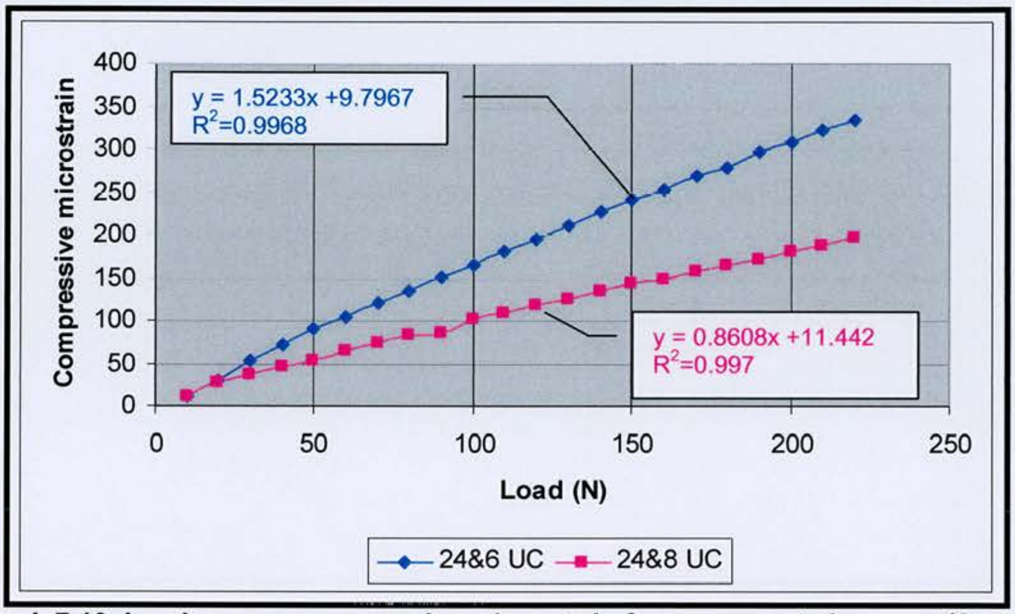
Materials and method:

Five nickel chromium dies of 24° TOC and 8mm axial wall height were produced as described in Section 6.1.1. On each new die, a new gold crown was fabricated with a thickness of 0.5mm occlusally and axially. The crowns were produced as described in Section 6.2. Two strain gauges with their soldering terminals were installed as described in Section 6.3. Each crown was in turn tested uncemented, partially cemented and fully cemented. The cementation procedure was described in Section 6.4. Using the Instron testing machine, an incrementally increasing compressive load was applied. The load magnitude was from 0N up to 220N in 10N increments. These cycles were repeated five times for each sample and the resultant strain was recorded and manipulated. The loading protocol was described in Section 6.5.2.

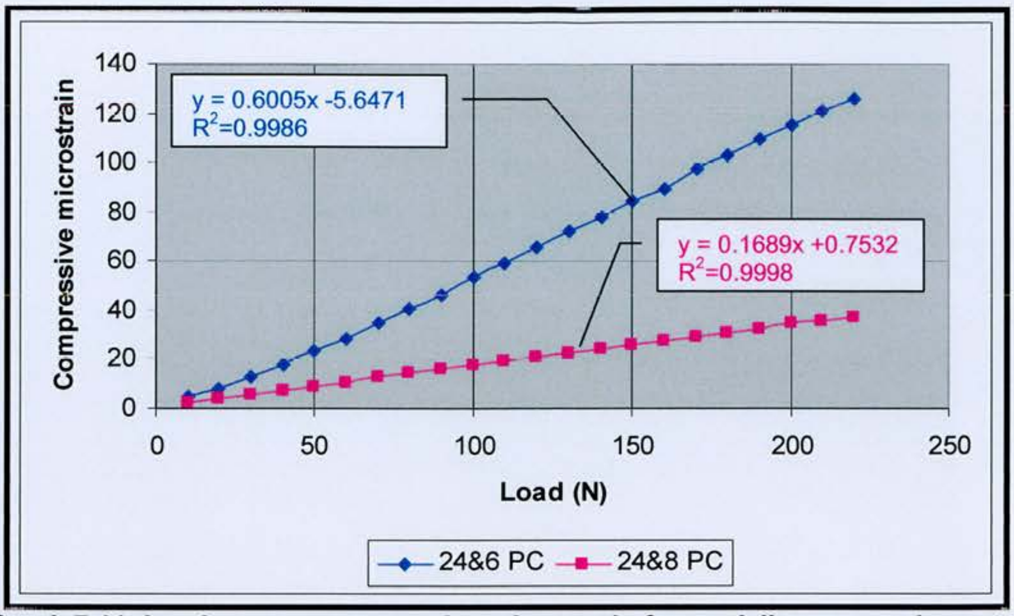
Results:

The results of loading (Newtons) were recorded and plotted against mean compressive strain (microstrain) for the 24° TOC and 8mm axial height crowns and those with 24° TOC but of 6mm axial height.

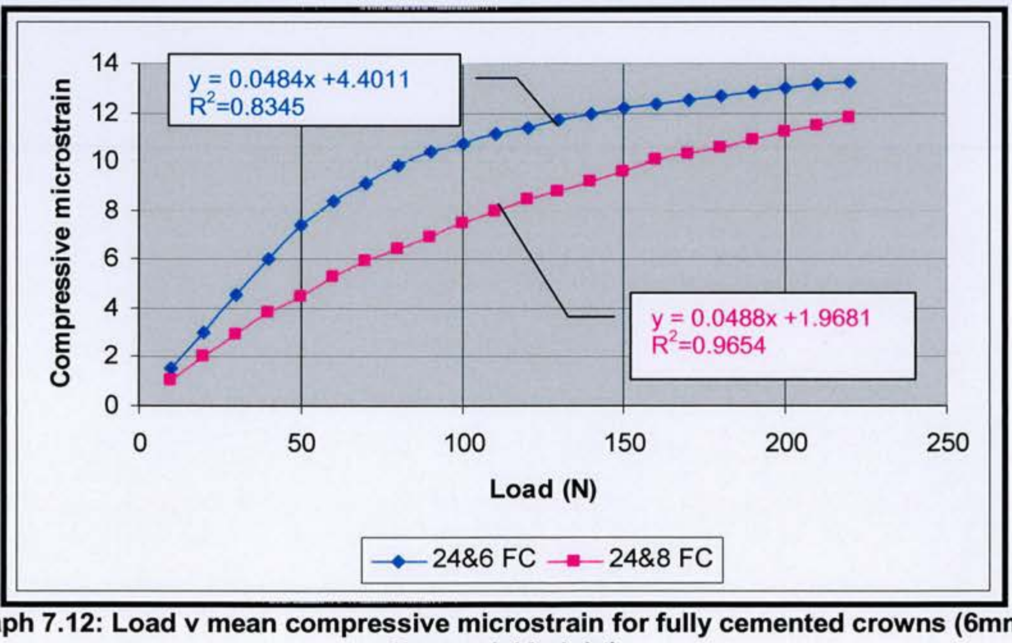
The results were shown in Graph 7.10 (uncemented), Graph 7.11 (partially cemented), and Graph 7.12 (fully cemented).



Graph 7.10: Load v mean compressive microstrain for un-cemented crowns (6mm & 8mm axial height).



Graph 7.11: Load v mean compressive microstrain for partially cemented crowns (6mm & 8mm axial height).



Graph 7.12: Load v mean compressive microstrain for fully cemented crowns (6mm & 8mm axial height)

Discussion:

Graph 7.10 (note the scale for Y axis) showed greater mean microstrain for 6mm axial wall height compared with 8mm axial wall height for the uncemented crowns. The highest value for 6mm was approximately 340 microstrain whilst for 8mm was 200 microstrain.

Graph 7.11 (note the scale for Y axis) showed the graphs for the partially cemented crowns. This showed that the highest mean microstrain for 6mm axial wall height was approximately 125 microstrain and for 8mm was about 40 microstrain.

Graph 7.12 (note the scale for Y axis) for the fully cemented crowns showed very low values compared to the partially and fully cemented samples. For 6mm axial wall height, it was about 13 microstrain whilst for 8mm axial wall height, it was about 12 microstrain.

The strain magnitude was less for 8mm axial wall height compared with 6mm with the same TOC (24^{o}) for all cementation conditions. Strain in the uncemented crowns was higher than for the partially cemented crowns, which in turn was higher than the fully cemented castings. The slope of the strain for the fully cemented crowns (8mm axial height) was almost similar to that 6mm axial height.

The results showed that as the load increased, there was an increase in the strain on the axial surfaces of the gold crowns. Increasing axial wall height from 6mm to 8mm decreased the strain under all cementation conditions.

For the uncemented and partially cemented crowns, there was a linear increase in the resultant strain with the applied load. For the fully cemented crowns however, the graph showed a curved rather than a linear relationship. There was a gradual increase in the difference as the load increased the strain of both 6mm and 8mm increased until 100N where the difference between them started to decrease gradually. For the fully cemented crowns, the difference between 6mm and 8mm axial height was not as big as the uncemented and partially cemented samples. Additionally the graph did not show completely linear results. As the strain increased, the difference between them increased till about 100N then returned again to become similar again.

It was clear from the results in graphs 7.10, 7.11, and 7.12 that with the increase the axial height, there was a decrease in strain. This might have been due to the increase in the surface area between the tooth preparation and the restoration fitting surface associated with an increase in the area of cement. The surface area of the frustum with 24° taper and 8mm was 213.336mm^2 whilst the surface area for 24° taper and 6mm was 168.198mm^2. With the increase of the surface area, the load per square surface area will be less resulting in less strain. It might be that the luting cement would be more durable compared with the shorter preparation. This data indicated that maximum crown height would be advantageous.

In agreement with this, a close relationship was found between axial height and retention by Kaufman *et al* (1961). They found that as the height increased, so did the retentive ability of the crown.

Conclusions:

1. The increase in the axial crown height decreased the strain on the axial surface of gold crown.
2. The presence of cement at the interface between the crown and the die reduced the strain in the axial walls of the crown.
3. The most strain was seen in the uncemented crowns and least in those which were fully cemented.
4. The effect of the extent of the cement film influenced axial wall strain more than the height of the preparation.

7.1.4. Experiments related to the crown casting

7.1.4.1. Effect of heat treatment and occlusal thickness of the crown on axial strain of fully cemented gold crown

Introduction:

There is scant information in the literature regarding the effect of heat treatment of cast gold crown and increased occlusal surface thickness on the axial strain.

Aims:

1. To determine the amount of strain on the axial surfaces of cemented gold crowns subjected to heat treatment.
2. To investigate the effects of heat treatment and occlusal surface thickness on the axial surface strain distribution.

Materials and method:

Five nickel chromium dies of 12° TOC and 6mm axial wall height were used for the experiment. For each die, three spaced gold crowns were cast as described in section 6.2 to produce fifteen gold crowns. The first ten gold crowns were cast with an overall thickness of 0.5mm; five of them were heat-treated and the other five were used as control samples. The last five gold crowns were fabricated with thickness of 0.5mm axially with occlusal thickness of 1.5mm.

The heat-treated five gold crowns that had 0.5mm overall thickness were placed in a porcelain furnace (Multimat ® Mack III, Dentsply) and fired in air. The temperature was increased to 600ºC and then reduced to 200ºC over 20 minutes before quenching under cold tap water. The external surfaces of the crowns were re-polished to remove the black discoloration caused by the heat treatment. The procedure for hardening heat treatment of the gold crowns was based on McCabe and Walls (1998).

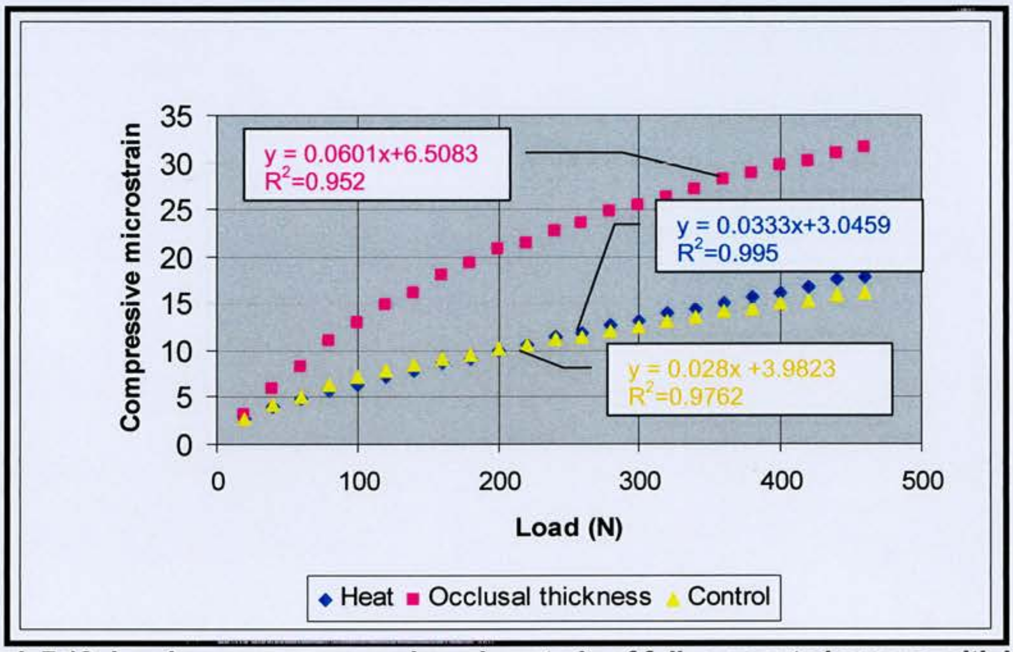
All the samples were fully cemented with the fitting surface of the crowns half filled with zinc phosphate cement as described in Section 6.4.

The Instron testing machine applied load through a new flat-ended steel applicator with a diameter of 6.35mm. The load application in this experiment was up to 460N and was increased incrementally by 20N. The loading was repeated five times for each sample. In all other respects, loading followed the method described in Section 6.5.

The compressive strain was recorded against applied load in Newtons as described in Section 6.6.

Results:

The results were shown in Graph 7.13 with mean compressive strain shown as positive values.



Graph 7.13: Load v mean compressive microstrain of fully cemented crowns with heat treatment and increased occlusal surface thickness.

Discussion:

The heat treatment of some gold crowns in the experiment was based on McCabe and Walls (1998). The procedure of hardening heat treatment was designed for gold type 3 and 4 alloys and was performed in pyrometrically controlled furnace. Additionally, the castings were supported by refractory material based on the recommendations of McCabe and Walls (1998).

The results showed that whilst there was no difference between the strain response for the control and the heat-treated samples, the strain was approximately doubled when the occlusal thickness increased from 0.5mm to 1.5mm. This might have been either related to the modulus of elasticity of the material that is not affected by heat, or this increase in the strain value with occlusal thickness increase might have been related to the increased rigidity of the crowns or increased thickness led to increase in the distance from the surface to the crown margins, providing increased leverage. However, the strain gauges were related to the margins not to the occlusal surface.

In the case of heat treatment and no heat treatment, the lack of difference might have been because heat treatment has no effect on the modulus of elasticity. Whilst the increased strain recorded for the crowns with thicker occlusal surfaces was probably the result of the increased rigidity of the crowns. This would have caused increased strain to be transmitted to the axial surface of the crown.

In the experiment using crowns of increased occlusal thickness, less axial strain was found with the thinner occlusal surface. This data should be interpreted with care. One conclusion could have been because the thinner occlusal surface produced less axial strain, such crowns might perform better clinically. However, it was likely that the thinner crowns deformed occlusally more readily than the thicker ones leading to less stress transfer to the axial walls. The effects of this on a thin crown cemented clinically would be likely to be adverse.

The trend of the increased occlusal thickness strain from the graph did not show entirely linear progression, instead it curved down at approximately 200N.

The increased compressive loading from 220N to 460N was a preliminary step for the dynamic loading. Additionally, instead of the pointed head of the upper jaw of the Instron machine, a flat steel tip with 6.35mm in diameter was used to minimise the likelihood of deformation of the occlusal surface of the cemented gold crown. Any permanent deformation would have changed the integrity of the underlying cement layer and might have affected the strains recorded. However, the flat surface of the Instron head would only apply a truly axial load if the crown had seated on cementation without skewing.

For a full gold crown, occlusal reduction should be of the order of 1.0-1.5mm (Shillingburg *et al* 1997). However, increasing the occlusal reduction clinically and allowing greater thickness of the casting occlusally leads to a reduction in axial wall height. This in turn would increase strain in the axial walls possibly negating any benefit produced by the more rigid occlusal surface.

Conclusions:

Heat treatment of gold crowns did not affect the strain measured in the axial walls of the cemented crowns. However, increasing the gold thickness on the occlusal surface from 0.5 to 1.5mm affected the strain distribution by virtually doubling the strain on the axial surfaces of the cemented gold crowns.

1. Heat-treatment of crowns with 0.5mm occlusal thickness did not alter the values recorded for axial strain.

2. The crowns with increased occlusal thickness (1.5mm) demonstrated higher levels of axial strain when compared with crowns of 0.5mm occlusal thickness with and without heat-treatment.

7.2 Dynamic loading

Introduction:

A cast restoration cemented in patient's mouth is exposed to repetitive loading over a prolonged period of time. The oral environment has high relative humidity. Dynamic loading experiments were considered an essential part of the investigation as being more clinically representative.

The cemented restoration *in vivo* is surrounded by aqueous solutions. The luting cement must be able to resist dissolution and disintegration that otherwise could lead to failure of the crown. From the literature, it was confirmed that humidity had an important role in the behaviour of dental luting cements and might have been an important factor in the survival of a cemented crown.

Using the dynamic loading experiments with the scanning electron microscope (SEM) allows the direct assessment of the affects on testing of the cement film.

Objectives:

1. To develop a method of assessing indirectly the bonding provided by the cement of the casting to the die under a simulated physiologic loading.
2. To examine the effect of uni-axial loading using a repeated known number of cycles on partially cemented cast gold crowns on cast nickel chromium dies.

3. To determine the effect of different aqueous media under dynamic loading.
4. To examine the cement film after dynamic loading directly using scanning electron and light microscopy.
5. To compare the data from the dynamic loading with observations of the cement film following testing.

Materials and method:

Eight nickel chromium dies of 12^{o} degrees TOC and 6mm axial wall height with 11mm diameter were used as described in Section 6.1. The dies were received two coats of die spacer occlusally and axially. The crowns were made of the same gold alloy with overall thickness of 0.5mm axially and occlusally and all the fabrication techniques followed those described in Section 6.2. Two strain gauges were used for each sample, installed opposite each other on the axial surfaces of the crowns 1mm above the crown-die junction margin as described in Section 6.3.

All the samples were partially cemented with zinc phosphate cement (De Trey®Zinc, Dentsply Ltd.). The crowns were placed on their respective dies and oriented in the same way. All the samples were left 24 hours to set. The procedure was described in Section 6.4.

The dynamic fatigue testing was carried out using the Instron testing machine as described in Section 6.5 but with instructions set for dynamic tests. The testing procedure was carried out by adjusting the Instron Universal Testing machine for cyclic compression loading on the samples for up to about $300X10^3$ cycles for each sample with frequency of the order of 2 cyles /second. The load magnitude was varied between 0-450N.

The initial position of the machine was zeroed. This position was taken as zero load magnitude from which the machine compressed the sample up to 450N and returned back to 0N to complete one cycle.

To control and stabilise the position of the samples on the lower jaw of the Instron machine and prevent them from lateral displacement, a metal adaptor was fabricated which accommodated the die base. This located the die ensuring that the loads were applied in the long axis of the crown and die.

For each testing condition, two samples were used.

Data from the two strain gauges was recorded for the duration of the test period following the procedure described in Section 6.6. Maximum, minimum, and mean values for microstrain were determined for each cycle and recorded by the software as described in Section 6.6.

The amplifiers were re-calibrated for the dynamic testing. The calibration graph for dynamic loading was shown in Graph 6.2. The calibration factor for amplifier 1 was 1.192 and amplifier 2 was 1.095.

The dynamic loading experiments were divided into the following:

1. Dynamic loading of dry partially cemented crowns.
2. Dynamic loading of wet partially cemented crowns

These were further subdivided into:

   2.1. Dynamic loading of partially cemented crowns in water

   2.2. Dynamic loading of partially cemented crowns in water and acid.

   2.3. Dynamic loading of partially cemented crowns in acid.

3. All the previous four categories were examined visually for the cement lute after being sectioned into two halves.

7.2.1. Fatigue testing of cemented assemblies under dry condition:

Materials and method:

Two samples were cemented and stored dry for at approximately 24 hours before testing as described in Section 6.4. The crowns were tested dry under dynamic loads from 0N up to 450N as described in Section 6.5.

Results:

The results were displayed graphically, as plots of the number of cycles against strain (microstrain, μ). Each graph for each sample showed the maximum, minimum and mean values of microstrain for each cycle.

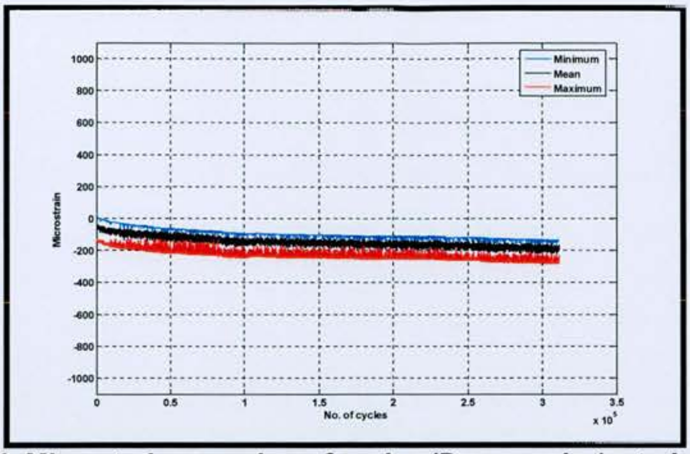
The results of samples that underwent fatigue tested under dry condition were shown in Graphs 7.14-7.17.

Graph 7.14 showed microstrain values with number of cycles for strain gauge 1 (sample 1).

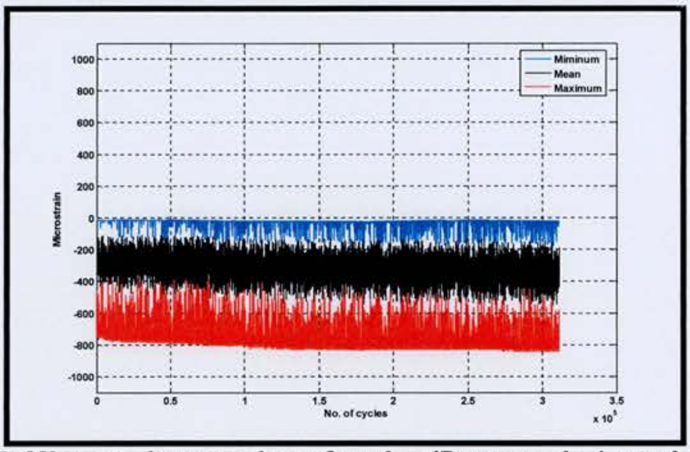
Graph 7.15 showed microstrain values with number of cycles for strain gauge 2 (sample 1).

Graph 7.16 showed microstrain values with number of cycles for strain gauge 1 (sample 2).

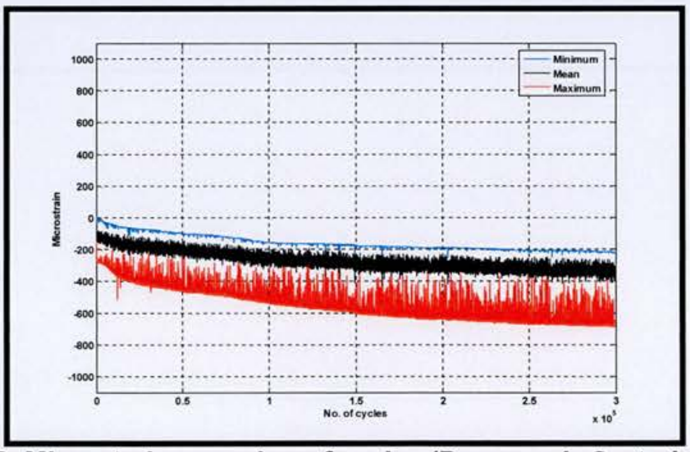
Graph 7.17 showed microstrain values with number of cycles for strain gauge 2 (sample 2).



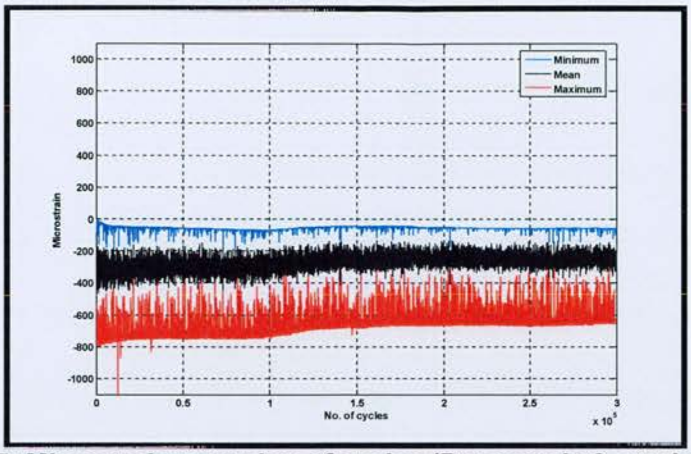
Graph 7.14: Microstrain v number of cycles (Dry sample 1, strain gauge 1).



Graph 7.15: Microstrain v number of cycles (Dry sample 1, strain gauge 2).



Graph 7.16: Microstrain v number of cycles (Dry sample 2, strain gauge 1).



Graph 7.17: Microstrain v number of cycles (Dry sample 2, strain gauge 2).

Discussion:

The mean values displayed were derived from the calibrated actual readings. Thus negative values represent increased compressive strain.

For both samples the graphs generally showed a continuous gradual increase in the compressive strain in one gauge with almost stable strain in the opposing one.

For sample 1, the graph of stain gauge 1 (Graph 7.14) showed a slow, gradual increase in strain with an increasing number of cycles. The mean strain measurements increased by approximately four folds from -50 microstrain into -200 microstrain.

For strain gauge 2 (Graph 7.15) however, the graph showed that the mean strain remained almost constant over the test period, there was a slight increase in the mean strain from -300 microstrain into about -350 microstrain.

Likewise, the graph for strain gauge 1 of sample 2 (Graph 7.16) showed an increase in the mean strain from -100 microstrain into about -300 microstrain with three folds increase.

For strain gauge 2 (Graph 7.17), the graph showed a slight decrease in the strain from about -380 microstrain which was constant until the cycle number of about 100,000 cycles into approximately -300 microstrain, which was also stable till the end of the experiment.

One interpretation of the difference in the graph from strain gauge 1 and 2 in the two samples was that there might have been movement on one side of the crown whilst the opposing side was stable. This might have explained the gradual increase in the strain recorded by one gauge with the virtual stability of the other. This strain trend could have been explained by a creep-like deformation of the cement film because of the repeated compressive stress. This could have produced a reorganisation of mechanical interlocking at the interface between metals and the cement. This observation had some support from the earlier work of Shen (1996) who was cited by Yamashita *et al* (2000).

This finding might also possibly have been caused by skewing of the crown during cementation or less likely that the load was not applied absolutely in the middle of the crown.

The strain gauge 2 was consistent as was strain gauge 1. It was possible that the amplifiers for the 2 strain gauges behaved differently. However, this was unlikely as the same amplifiers had been used for the re-calibration described in Material and Method Section 6.6.1.

A further possibility was the influence of the finish line of the preparation relative to the margin of the casting. It was possible that in the marginal area related to strain gauge 1 that the casting moved whilst in the area of gauge 2, the margins on the die and the casting more effectively resisted apical movement. Again, this seemed improbable as the dies were uniform and the crown made using identical methods. It would also have been an unusual coincidence if this phenomenon had occurred in the area of strain gauge 2 on both occasions.

The mean values for microstrain showed small and gradual changes. Comparing with the data for the uncemented, partially and fully cemented states showed in Graph 7.4 indicated that neither of these samples in this series of the experiment showed any marked tendency to mimic the uncemented crowns.

The literature did support the concept of the finishing line geometry influencing strain distributions. Kamposiora *et al* (1994) reported using 2 dimensional finite element strain analysis that there was almost no difference between a chamfer and shoulder marginal configuration except at the edge of the margin where the chamfer finish lines produced 2 to 8 times greater stresses. This was considered unlikely to have been the reason for the observed differences in the experiment.

Conclusions:

1. Application of increasing dynamic loads to the occlusal surfaces of the partially cemented crowns produced increasing axial strain. The strains were more marked on one side compared with the other.
2. The graph indicated no evidence of failure of the cement lute during testing of the dry specimens.

7.2.2. Dynamic loading of wet partially cemented crowns:

7.2.2.1. Dynamic loading of partially cemented crowns in water.

Materials and method:

Two crown-die assemblies were used in this part of the study as described in Section 7.2.1. Additionally, the strain gauges were coated with a high performance room temperature curing silicone sealant (Dow Corning®744 sealant) which was left for 24 hours to cure before being the samples tested. A fluid container was constructed from regular addition silicone impression material (Provil®novo, Heraeus Kulzer Gmbh&Co. KG. Gruner Wer 11 . D-634550 Hanau, Germany) and was used to surround the die and crown. This was shown in Figure 7.14.

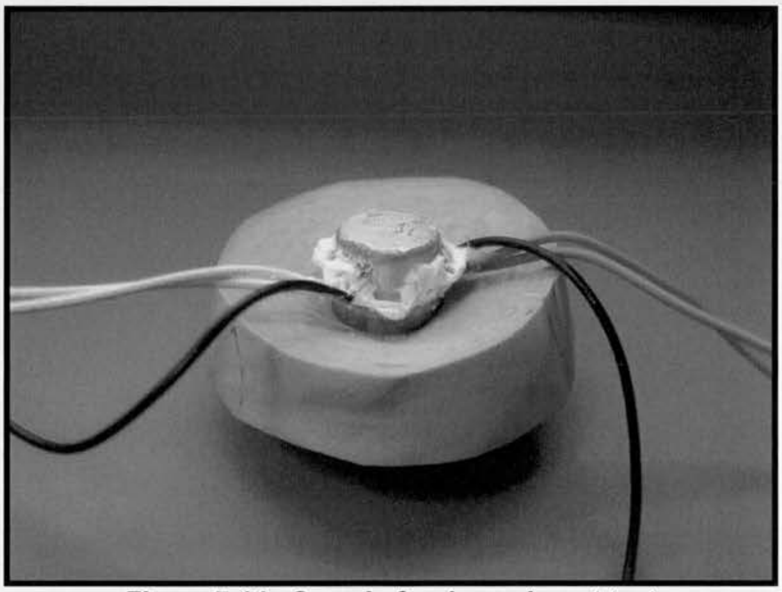


Figure 7.14.: Sample for dynamic wet test

The silicone container was approximately 10cm wide. It extended from about 1cm below the crown margin onto the trunk of the dies. It was shaped to hold fluid and allow coverage about 3mm of the axial surface of the gold crowns.

The silicone containers were glued to the stump of the die using Cyanoacrylate Kwikfix Super Glue (Chemence LTD, Princewood Road, Corby, Northants, NN17 4XD).

The crowns were loaded and the data acquired and recorded as described in Section 6.6.

Results:

The results were displayed graphically. The results showed the number of cycles against strain (microstrain). Each graph for each sample showed the maximum, minimum values of strain and the mean reading for each cycle.

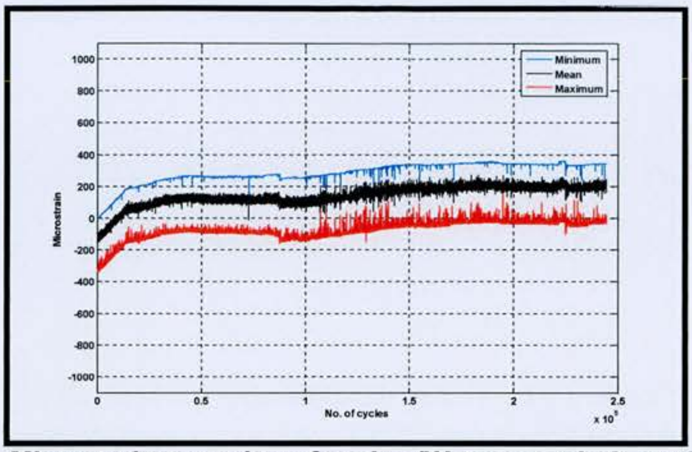
The results of samples that underwent fatigue tested under water during all the testing period were shown in Graphs 7.18-7.21.

Graph 7.18 showed microstrain values with number of cycles for strain gauge 1 (sample 3).

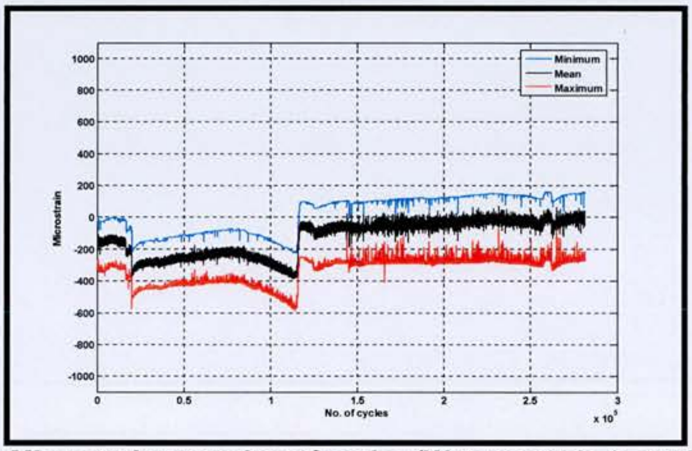
Graph 7.19 showed microstrain values with number of cycles for strain gauge 2 (sample 3).

Graph 7.20 showed microstrain values with number of cycles for strain gauge 1 (sample 4).

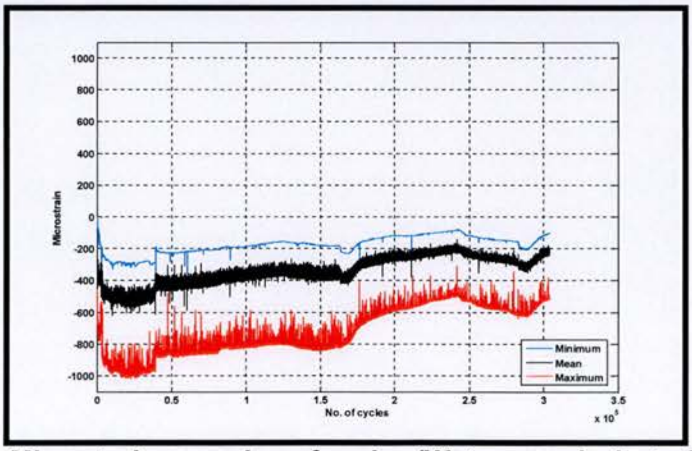
Graph 7.21 showed microstrain values with number of cycles for strain gauge 2 (sample 4).



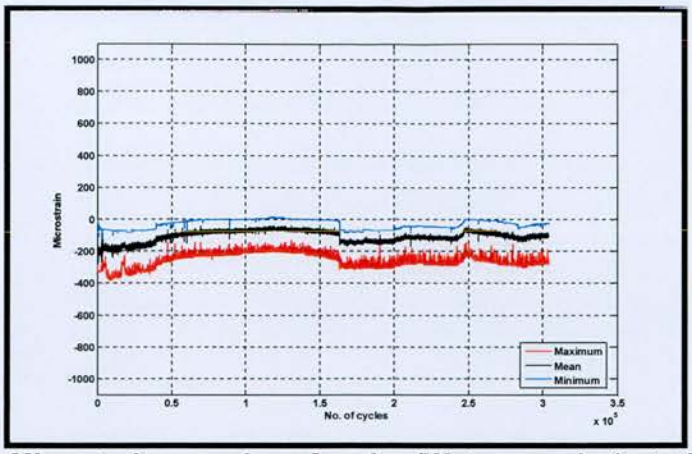
Graph 7.18: Microstrain v number of cycles (Water sample 3, strain gauge 1).



Graph 7.19: Microstrain v number of cycles (Water sample 3, strain gauge 2).



Graph 7.20: Microstrain v number of cycles (Water sample 4, strain gauge 1).



Graph 7.21: Microstrain v number of cycles (Water sample 4, strain gauge 2).

Discussion:

The trend for the samples tested under water showed a gradual trend towards a decrease in the compressive strain. This was unexpected.

Sample test 3, the gauge 1 (Graph 7.18) showed a steep progressive decrease in mean compressive strain from -100 to about 100 microstrain from the start of the cycling up to about 14000 cycles, after that, the mean strain decreased more slowly from 100 to about 200 microstrain.

Strain gauge 2 (Graph 7.19) showed an initial value of about -190 microstrains until about 20000 cycles when it increased sharply to about -250 microstrains. The mean strain then decreased to about -200 microstrains at about 75000 cycles before showing a further decrease to about -50 microstrains at about 120000 cycles. The mean strain then decreased gradually to about 0 microstrain until the end of the cycling.

There was a point of similar behaviour of the two strain gauges at about 270000 cycles when the mean strain decreased sharply.

The mean strain trend of the two strain gauges of sample 4 (Figure7.20, Figure 7.21) showed a decrease in the mean compressive strain from the start of the cycling to the completion for two gauges.
Strain gauge 1 trend (Graph 7.20) showed a sharp increase in the mean strain to -400 microstrain, then increased to about -550 microstrains. The mean strain decreased gradually till about 40000 cycles, then decreased sharply to approximately -450 microstrain. The mean strain then decreased gradually to about -200 microstrain at the end of the cycling. Toward the end of testing, there were two sudden changes; at the about 170000 and 270000 cycles, the mean strain increased to about -400 microstrain at 17000 cycles. At 24000, the mean strain trend increased to about -300 microstrains before increasing again sharply to -350 microstrain at about 280000 cycles. The mean strain then decreased again until the completion of the cycling.
The strain gauge 2 (Graph 7.21) showed a mean decrease from -225 to about -190 microstrain. There was a sharp decrease in strain at the start of the cycling around 40000 cycles. At that point, the trend decreased reaching a maximum of approximately -100 microstrain. It then increased gradually to reach -125 microstrain. There was a sudden increase in the mean strain at

approximately 160000 cycles to about -180 microstrain; the mean strain then decreased gradually again until 240000 cycles to record about -150 microstrain. The mean strain decreased sharply to about -110 microstrain, then increased sharply again to -120 microstrain at 270000 cycles. The mean strain then decreased to -140 microstrain at the end of cycling.

There was a similarity between the data from the two strain gauges. There were multiple sharp changes in the values recorded by the two strain gauges; these might have represented a possible change in the cement film affecting the support of the gold crown. Comparing the data with those obtained from the dry specimens showed that the hydration of cement affected the strain in the axial walls of the crowns. The overall trend was for the strain to have reduced; this was not indicative of any deterioration in the bond between the crown and the die. Rather, what was observed might have represented the difference between a dry and one that was hydrated.

These samples did not behave in any way like the uncemented, partially cemented and fully cemented crowns shown in Graph 7.4. Rather they showed decreasing compressive strain tending apparently to more towards a state where the bond provided by the cement was more extensive.

Comparing with other studies, the difference might have represented dimensional changes in the cement film after water sorption. It has been confirmed that zinc phosphate cement is sensitive to water. *In vitro* studies showed that dehydration of water-based cements results in cracking which represents a cohesive failure in the cement material (Mount, 1991). Watson *et al* (1998) reported that these cracks could be found on the surface and at the tooth-restoration interfaces. Upon addition of water however, the cement partially or completely recovered its initial volume.

Oilo (1975) described a method of measuring the linear dimensional changes during the setting of zinc phosphate cement. Measurement started 3 minutes after mixing at 37°C and under various environmental humidities. He reported that the contraction under dry conditions was more pronounced. Wilson *et al* (1979) showed that there was a linear relationship between the degree of hydration and the strength of dental luting cement. All the cements were found to become more highly hydrated and stronger as they aged. They related that to the hydration of the cement used, the weakest cement was the least hydrated cement and they hypothesised that the water acts as a plasticiser. They related the strength of dental cements directly to the ratio of non-evaporated water.

Hondrum (2000) investigated the effect of evaporation on the properties of water based luting cements; he concluded that mechanically zinc phosphate and resin glass ionomer cements were comparatively resistant to the effects of evaporation.

Conclusions:

1. There was a broadly linear correlation between decreasing strain on the axial walls and increasing dynamic loading.
2. Comparison with the dry specimens, hydration of zinc phosphate cement, showed a noticeable effect on stress-strain behaviour.

7.2.2.2. Dynamic loading of partially cemented crowns under water and acid.

Materials and method:

The method was as described in Section 7.2. Two samples were partially cemented and left to set for 24 hours. The cemented samples were then dynamically loaded between 0N-450N for about 300×10^3 cycles with frequency of 2Hz. The cemented samples were then immersed in acid during day and distilled water during night.

For the samples that were tested under acidic media, 0.1M lactic acid solution (0.1M acid) and 1M sodium lactate/lactic acid solution buffer solution (1M buffer) were chosen as the aqueous media.

The initial pH of the solution was 2.74. The solutions were prepared 24 hours before use and the pH was checked with a pH meter immediately before use. The solution specification was shown in Table 7.3.

Category	Lactic acid	Sodium lactate (General purpose grade)
Net	500 ml	500 ml
Code	L/0100/PB08	S/5161/08
Batch Number	0559904	0573165

Table 7.3: The specifications of the acidic solution

A lactic acid/lactate buffer with a pH of 2.74 was made. The pH of 0.1M lactic acid was corrected using 1M sodium lactate; 100 ml of 0.1M lactic acid required 9mls sodium lactate. The pH was checked on the morning before use and found to have remained constant.

Results:

The results were showed graphically and showed the number of cycles against strain. Each graph for each sample showed the maximum, minimum and mean values for each cycle.

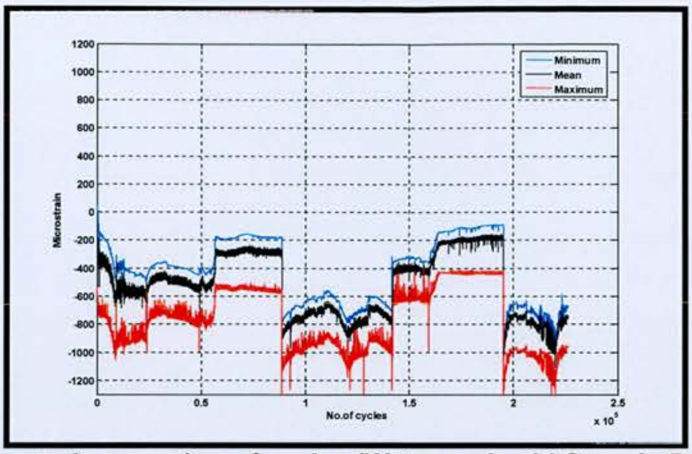
The results of samples that underwent fatigue testing under acid during the day and were kept in water overnight were shown in Graphs 7.22- 7.25.

Graph 7.22 showed microstrain values with number of cycles for strain gauge 1 (sample 5).

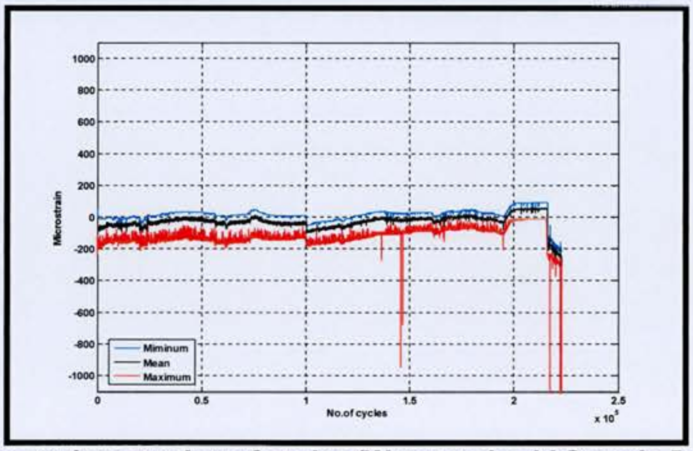
Graph 7.23 showed microstrain values with number of cycles for strain gauge 2 (sample 5).

Graph 7.24 showed microstrain values with number of cycles for strain gauge 1 (sample 6).

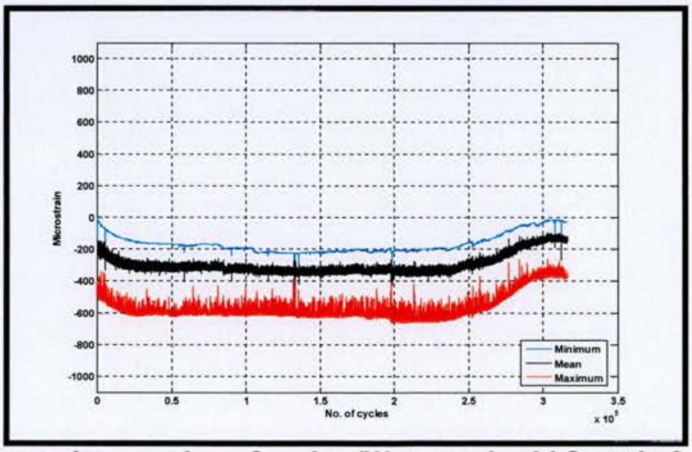
Graph 7.25 showed microstrain values with number of cycles for strain gauge 2 (sample 6).



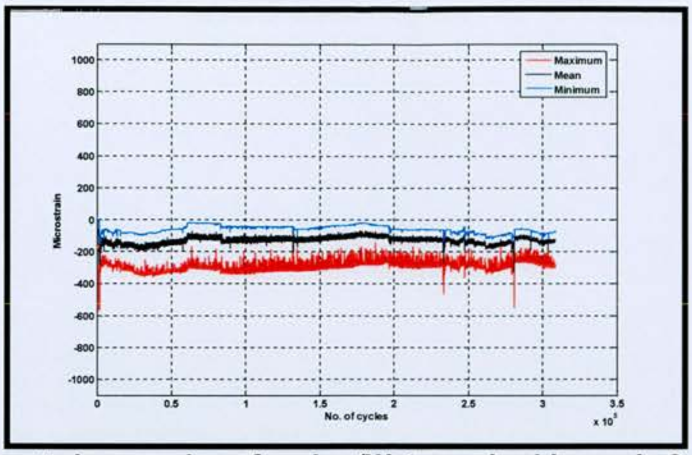
Graph 7.22: Microstrain v number of cycles (Water and acid Sample 5, strain gauge 1).



Graph 7.23: Microstrain v number of cycles (Water and acid Sample 5, strain gauge 2).



Graph 7.24: Microstrain v number of cycles (Water and acid Sample 6, strain gauge 1).



Graph 7.25: Microstrain v number of cycles (Water and acid sample 6, strain gauge 2).

Discussion:

Two samples were cemented and tested in acid and water alternately for 12 hours.

The graphs of mean strain against number of cycles showed a number of increases and decreases.

Strain gauge 1 of sample 5 (Graph 7.22) showed an overall increase in the mean strain over the testing period from approximately -300 to -900 microstrain. The increase in the strain was not gradual over the test period. It showed a number of sharp increases at irregular intervals. Following each rapid increase in the mean strain which varied in magnitude between about -400 and -1000, the strains either decreased gradually or remained constant.

The tracing graph of strain gauge 2 of sample 5 (Graph 7.23) showed a small decrease over the cycling period. The mean strain generally decreased from approximately -100 microstrain to about 80 microstrain.

It was noted that, there were parallel but opposite changes in the two strain gauges of this sample, where as the strain increased on one side of the crown, it decreased by about similar amount on the opposite side. This might have indicated movement of the crown on its die which in turn could have meant a failure in the cement film had occurred.

Gauge 1 of sample 6 (Graph 7.24) showed that the mean strain increased at about 20000 cycles from approximately -200 to -300 microstrain. There was a very small increase to -400 microstrain until about 240000 cycles, then the strain decreased sharply to -180 microstrain.

Regarding strain gauge 2 of sample 6, the graph (Graph 7.25) showed a mean strain about -180 microstrain over the entire cycling period. It was clear from the tracing trend; there was a decrease around 2.5×10^5 cycles which might have represented minute movement of the gold crown on its die reflecting a change in the integrity of the luting cement.

Hydration is necessary for optimal performance of zinc phosphate cement and this might apply to all acid based cements. The acid was damaging to zinc phosphate cement reflecting by the increased strain in the axial walls of the crowns; however the increase in the strain generally did not continue. The graphs showed that there was a tendency for the strain to decrease again after these brief increases.

The changes recorded by strain gauge 1 in Sample 5 (Graph 7.22) were marked by comparison to the other gauges. The sudden marked increases in compressive strain might have indicated deterioration in part of the support provided for the crown by the cement. The values increased greatly and appeared to follow the behaviour recorded by the uncemented crowns in the static loading experiments (Graph 7.4).

The cement matrix might have been damaged by the acid leading to a reduction in its effectiveness. The cement then appeared to re-establish its ability to support the crown on the die and the strain either remained relatively constant or diminished slightly. It did not seem logical that the cement could have re-bonded the crown to the die and the analogy proposed by Taggart (1907) that disintegrated cement could act like sand and block or jam the crown-die interface was an interestingly possible.

Conclusions:

1. The introduction of acid into the aqueous medium resulted in generally increased axial strain.
2. Immersion of the cemented crown in acid made their response to axial loading more variable.

7.2.2.3. Dynamic loading of partially cemented crowns under acid.

Materials and method:

The method was as described in Section 7.2.2 and Section 7.2.2.2 for lactic acid solution preparation. Two samples were partially cemented and left to set for 24 hours. The cemented samples were then dynamically loaded between 0 and 450N for about $300x10^3$ cycles with frequency of 2Hz. The cemented samples were then immersed in acid during day and night.

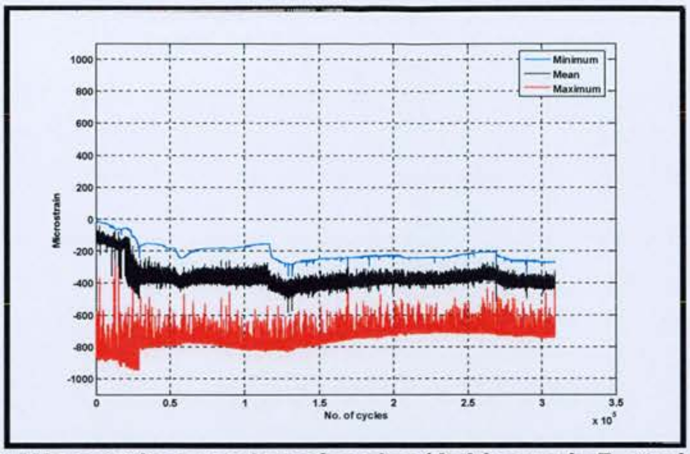
Results:

The results were shown graphically as number of cycles against strain (μ, microstrain). Each graph for each sample showed the maximum, minimum, and mean values of strain for each cycle.

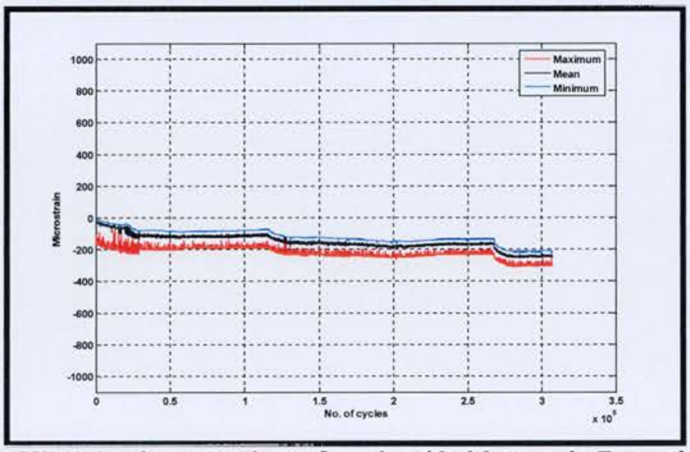
The results for the two samples were shown in Graphs 7.26-7.29.

Graph 7.26 showed the microstrain for strain gauge 1 (sample 7). Graph 7.27 showed the results microstrain for strain gauge 2 (sample 7).

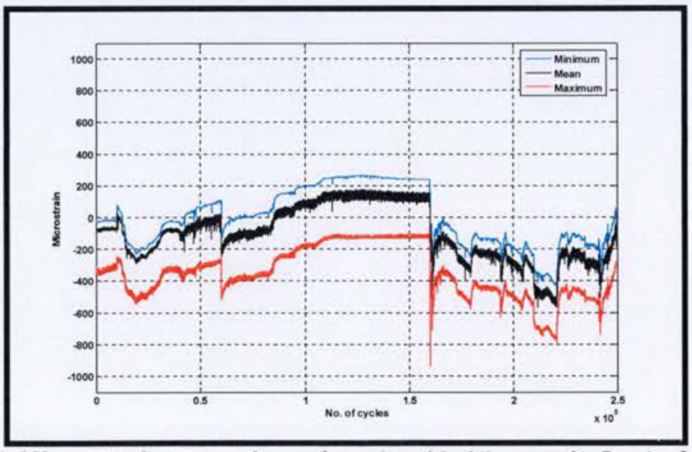
Graph 7.28 showed the microstrain for strain gauge 1 (sample 8). Eventually, Graph 7.29 showed the microstrain for strain gauge 2 (sample 8).



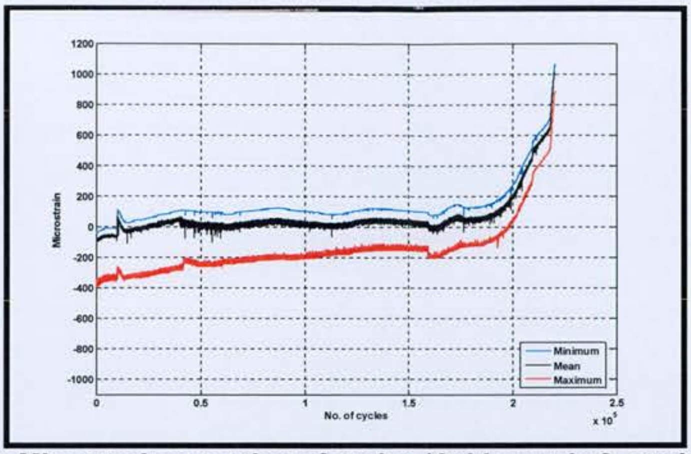
Graph 7.26: Microstrain v number of cycles (Acid sample 7, strain gauge 1).



Graph 7.27: Microstrain v number of cycles (Acid sample 7, strain gauge 2).



Graph 7.28: Microstrain v number of cycles (Acid sample 8, strain gauge 1).



Graph 7.29: Microstrain v number of cycles (Acid sample 8, strain gauge 2).

Discussion:

In this experiment, the samples were immersed in lactic acid buffer solution of pH 2.74 throughout the test period. The recommendation for the acid solution was based on Nomoto and McCabe (2001).

The general trend for these samples immersed in acid was for rapid changes in strain as the number of cycles increased. The changes were not constant but sudden and at irregular intervals. The changes seemed to occur at the same time in opposing strain gauge but their magnitude was different. The data from three of the gauges showed an overall increase in compressive microstrain, whilst the fourth showed a decrease.

For strain gauge 1 of sample 7 (Graph 7.26), the mean microstrain increased suddenly at interval periods from approximately -100 to -500 microstrain.

Likewise strain gauge 2 (Graph 7.27) showed a sudden sharp increases in the mean strain with time. It increased to about -250 microstrain.

In the second sample (8), the Graph 7.28 for strain gauge 1 showed that the mean compressive strain increased from about -50 to about -500 microstrain. The trend was not stable during the cycling period with some increases and decreases.

The tracing might have been explained as being due to a disintegration of part of the cement film underneath. The sharp increases in compressive microstrain were indicative of the support for the crown by the cement being compromised. Compared with (Graph 7.4) they reflected a tendency for the crown to be relatively less effectively cemented. The traces further indicated periods when the cement appeared to become more effective when the compressive microstrain remained more constant or even decreased. The reasons for this were discussed in Section 7.2.2.2. There was no information which permitted further inferences to be drawn regarding the quality of the cement film.

Strain gauge 2 of sample 8 (graph 7.29) showed a marked but progressive decrease in the mean compressive strain from about -100 microstrain to more than (+) 600 microstrain. This previously unseen behaviour might have represented a skewing of the crown after the cement had been affected by the acid. The sample 8 graphs showed a dramatic change in strain after the addition of lactic acid. This was likely to have been the results of the acid damaging the cement film loading to deterioration in the bond between the crown and the die.

The need for direct investigation after dynamic loading raised the question as to whether strain gauge usage alone was able to determine the true behaviour of cement film. There was a need to do a direct invasive to look at the cement directly. There was a need to scanning electron microscope in the next experiment.

Conclusions:

1. The presence of the acidic medium had a marked effect on the axial strain recorded in the crowns.
2. There was a trend in 3 out of the 4 gauges for the compressive axial strain to increase. These showed a tendency to be moving toward the uncememted state.
3. The fourth strain gauge showed a response which indicated that the cement film was becoming more effective or complete.

7.2.3. Examination of the cement film using scanning electron microscope.

Materials and method:

Following dynamic testing, the crowns on their respected dies were embedded in a clear resin and sectioned as described in Section 6.7.

The samples sectioned and examined were:

1. Two samples cemented and tested dry, Section 7.2.1.
2. Two samples tested and stored in water, Section 7.2.2.1.
3. Two samples tested under acid and stored in water, Section 7.2.2.2.
4. Two samples tested and stored in acid, Section 7.2.2.3.

The sectioned samples were viewed under the scanning electron microscope (SEM 505 Philips) at magnifications between 1.2×10^3- 2.5×10^3 times. A protocol was followed and the scanned surfaces were photographed to record the fitting surface of the gold crown and the surface of the die beneath the mid-point of the strain gauges.

The photomicrographs were scanned and stored digitally. The photomicrographs of the sectioned crowns and dies were evaluated qualitatively and voids, fissures and evidence of apparent loss of attachment of the cement to the crown and for die were recorded. The procedure was described in Section 6.7.3.

Results:

A typical micrograph from each group was shown in Figures 7.15, 7.16, 7.17 and 7.18. Figure 7.15 for a dry sample, Figure 7.16 for a water sample, Figure 7.17 for water and acid samples and Figure 7.18 for an acid sample.

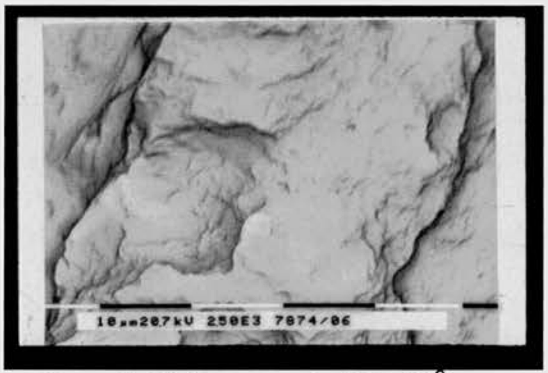


Figure 7.15: Dry sample (2.5x10^3)

Figure 7.16: water sample (1.31x10^3)

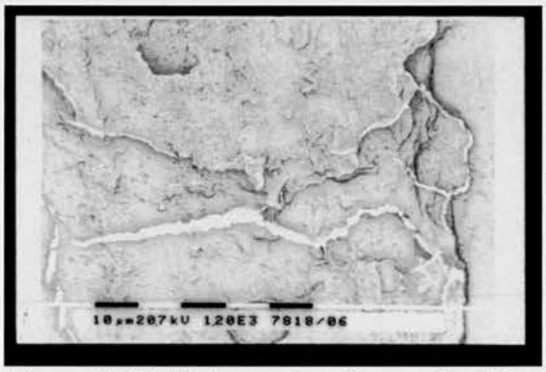


Figure 7.17: Water and acid sample (1.2x10^3)

Figure 7.18: Acid sample (1.93x10^3)

Discussion:

The process of embedding and sectioning might have changed the cement film. Light cure dental adhesive following sectioning was applied to minimise any further changes in the cement film and to minimise dehydration.

The scanning electron micrographs of dry samples showed that the cement film had few cracks and some minor voids were seen on the gold crown side. The cement film appeared broadly intact.

The sectioned crowns which had been tested in water showed more cracks of different sizes running in different directions. Some cracks were large with big pieces of cement between them. Cracks were also noted apparently separating the cement from both the die and the gold crown. However in some places the cement appeared intact. The crack size reached about 10 microns in width and extended from the crown fitting surface to the die surface. They were distributed along all the surface of the crown, top, middle and margins. There were no big losses of the cement at the margins however.

For the two samples tested in acid during the day and stored in water at night showed many cracks of different sizes, directions, and locations. Some cracks appeared to separate the cement from both sides die and crown. The cement film appeared badly damaged. There were places of severe destruction near the irregular sharp points of the fitting surfaces and the gold crown and the die.

The scanning electron micrographs of the two samples tested wholly in acid showed multiple cracks in the cement film running in different directions horizontally, vertically and obliquely. The size of these ranged from minor hair-like cracks to huge defects with the cracks being more evident near the margins. Between these cracks there were multiple intact isolated islands of cement material. Near the margins, the cement material became more granular showing a honey comb appearance. In extreme cases there was complete loss of the cement material from the interface between the crown and die near the margin, leaving it vacant. The cement film showed major changes.

Comparing the four groups, the cement film in the dry specimens appeared generally intact whilst the first changes in the appearance of the cement film were seen in those samples tested in water. The addition of acid as the aqueous medium for testing with storage of the samples in water at night further increased the apparent damage to the cement film. When the crowns and dies were both dynamically loaded in acid and stored in acid, the damage to the cement film was most sever. This was practically evident in the marginal areas where the cement had frequently entirely disappeared. The deterioration of the cement film appeared to be more related to the presence of moisture rather than the application of the dynamic load. This was supported by the strain measurements from the axial walls of the crowns during the dynamic tests.

It had been considered that there might have been a direct correlation between the strain measurements and the appearance of the cement film beneath the middle of the strain gauges. However, this was not observed. Further consideration of the experimental model made such a hypothesis unlikely as whilst there were differences in the strain behaviour of the four groups of cemented crowns, these differences may not have been due to alterations in the bond provided by the cement directly beneath the strain gauges. It was likely that changes in the nature of the attachment of the crown to the die in areas more distant from the strain gauges would have affected the strain recorded.

Conclusions:

1. Sectioning enabled direct viewing of the cement film between the crown and the die.
2. The dry cement appeared least damaged following testing.
3. The presence of an aqueous medium in addition to the application of dynamic loading produced changes in the appearance of the cement which appeared damaged as was evidenced by voids, cracks and signs of detachment of the cement from the crown and die.
4. The damage of the cement was increased not only by hydration but greatly by the presence of acid with marginal loss of the cement being observed.
5. No direct correlation of the degree of damage to the cement film with the strain measurements was evident.
6. The storage medium produced more damage than the dynamic loading.

Chapter 8: General Discussion

8.1. Discussion of materials and methods

8.1.1. Fabrication of Nickel chromium dies

Human teeth were not used in these experiments because storage conditions and post-extraction time were thought to influence their mechanical properties, Carter *et al* (1983), Strawn *et al* (1996). Additionally, it would have been difficult to standardise the size and shape of prepared natural teeth. Instead, metal dies were used to avoid fracture or distortion of the die during testing and to facilitate comparisons. The rational for using base metal alloys for the dies was their high compressive strength which minimise deformation during the application of high loads. Nickel chromium alloy was used because of its hardness and abrasion resistance. Chan (1982), Lockwood (1984) and Cassidy (1985) reported nickel chromium to be a suitable die material. They reported that it was abrasion resistant allowing repeated cementation and removal of the crowns without damage to either the die or crown. The results of using a nickel chromium alloy, which had an elastic moduli in the range of 90-100GPa (Craig 1997), as a die simulating tooth structure should not be transferable to other substrates.

For standardisation and comparative purposes, the same master stone model and procedures were used to fabricate all dies and crowns. The dimensions of the die used in the study represented an average of molar complete veneer crown preparation described in the literature. The cross-sectional width of the nickel chromium dies at the gingival margins was 11mm. This measurement was based on the work of Wheeler (1958) who reported that the diameter of the mandibular first molar mesio-distally was 11mm and bucco-lingually 10.5mm.

A minimum of 6mm axial wall height was needed for the dies in order to allow appropriate positioning of the strain gauges; these were 4.3mm long and 5.8mm wide as described in Section 6.3.1. In posterior teeth, it is rare to find more than 6mm axial height (Yamashita *et al*, 1997). They added that axial wall height of 6mm was the minimum which allowed placement of the strain gauges. Their location of about 1mm above the margins was for standardisation purposes. Goodacre *et al* (2001) emphasised that the minimum occluso-cervical dimension of molars should be 4mm when prepared with 10° to 20° TOC. However, an axial wall height of 6mm was still considered to be clinically representative.

The central screw was used to remove the crown after cementation to allow the crowns to be removed whilst reducing the possibilities of crown distortion and to avoid strain gauge damage that could have been produced by grasping the external surface of the crown or levering underneath its margin. During cementation, the screw was turned to be flush with the flat occlusal surface of the metallic die to prevent luting cement from being pushed into the tapped hole. It would have made the screw to unseat the crown difficult to turn and could also have acted as an internal vent.

The dies were made from nickel chromium alloy and had no periodontal ligament, which is completely different from the clinical reality; hence the results should be interpreted clinically with caution. However, it was reported by Yamashita (1997) regarding the suspensory effect of the periodontal ligament that strain distribution on single crown between *in vivo* and *in vitro* measurements was similar under static loading.

8.1.2. Fabrication of gold crowns

Silver coloured die spacer was used as Oliva *et al* (1988) had reported difficulty of application of the gold coloured material as it clumped. They had also reported that the golden colour provided a coarser surface texture with more irregularities compared to the silver spacer. This was in agreement with Rieger *et al* (1987) who reported that gold material provided a rougher appearance than the silver spacer. The type of brush used was that supplied by manufacturer. Oliva *et al* (1988) founded that the use of No.1 sable brush resulted in significantly increased spacer thickness compared with the brush supplied by the manufacturer. They concluded that using the same die spacer and coating technique on a number of dies would not produce uniform film thickness.

The dip wax technique was used for forming the wax patterns of the gold crowns. It had been shown that this technique could give crown thickness of about 0.5mm thickness (Kovarik *et al* 1992). During gold crown fabrication, wax was applied directly onto the spaced metal dies. During the wax dipping, it was noted that the thickness of the wax varied according to the time of dipping in the molten wax. For this study, the dipping time was standardised between 7-8 seconds.

The gold crowns were waxed on metal dies. It has been shown that the adaptation of a crown formed from a pattern waxed onto a metal die was better than for a pattern waxed on a stone die replica of a preparation (Fusayama *et al* 1963). It also seemed sensible to make the crown by a direct waxing technique rather than introducing an indirect stage which would have been likely to affect the consistency of adaptation of the casting.

Two aims in this study were to control the thickness of the gold crown and the load on the luting cement under the crown since Sugita *et al* (2000) emphasised that as metal thickness was increased, the deformation of the crown decreased and the cement failure load increased.

8.1.3. Luting cement and cementation process

Zinc phosphate cement was used in the series of experiments. Diaz-Arnold *et al* (1999) reported that for more than a century zinc phosphate cement has been the most widely used and oldest luting agent, in spite of some well documented disadvantages. It serves as the standard with which newer cements can be compared. Additionally, Margerit *et al* (1996) stated that of utmost important is the long clinical record of this cement. Its inherent stability was reported in a study that analysed the chemical structure of zinc phosphate cement samples obtained from 2 to 43 years. The proven reliability of this cement validated its use in this investigation.

The mixing technique was as described by Eames *et al* (1977). In these series of experiments, a standardised proportioning and mixing technique was followed for all cementations.

This study had a strong focus on the cement distribution between the gold crown and its respective die. There were three scenarios, uncemented (test for samples without cement in between crown and die), partially cemented (cement only on the axial surface of the die) and fully cemented (the cement covered all the surface of the die). Full cementation produced a complete cement film between the fitting surface of the crown and the die.

The partially cemented crown-die assemblies were used for experimental purposes to determine the effect of the cement distribution on the strain distribution under the given experimental load. It served as a predictor of deterioration in the cement lute.

The experiment was held at room temperature of 22+/-1°C. Mesu (1983) reported that the compressive strength of zinc phosphate cement decreased by about 10% when the temperature increased to 37°C compared to that mixed at 23°C.

The pressure exerted during cementation was 5kg, which followed Jorgenson's (1960, a) recommendation, who claimed that forces above that level had no significant effect on reducing film thicknesses whilst maintaining pressures longer than one minute had no appreciable effect either.

8.1.4. Loading procedure

Long term service is an essential characteristic of crowns and fixed partial dentures. The literature has drawbacks as most of the studies are *in vitro* that more related to retention. The commonest loads used in those studies have been tensile and non-axially directed. This study used axially-directed compressive load to investigate the variables that affect the long term survival of crowns and bridges.

The loads were uni-directional and applied at one point on the crown. The direction of the applied axial loads used in these experiments was likely to have been favourable to the cement lute in contrast with those directed non-axially; the latter would have produced a cantilevering action with increased resultant strain. This is based on research into pulling a crown off a simulated prepared tooth (in the form of metallic or plastic dies) in a direction parallel to the long axis of the tooth preparation (retention) (Jorgensen 1955). However, Dodge *et al* (1985) stated that forces that tend to remove cemented restorations along their path of withdrawal are small compared to those tend to seat or tilt those restorations. It was concluded that true tensile forces seldom occur especially on a single unit crown in the oral environment, and when they do occur, they certainly do not compare with those produced by occluding or chewing.

The load character for static loading was incremental up to 220N for the experiments of the die geometries and cement film thickness. The maximum load magnitude was for comparative purposes. In the experiment of heat treatment and increased occlusal thickness experiment the Instron testing machine applied load through a new flat-ended steel applicator with a diameter of 6.35mm to minimise the likelihood of the indent formation and permanent deformation of the tested crowns and their dies during testing. There was increased from 220N to 460N. That increase was considered a small amount of load and little happened to the cemented casting at lower load, since the focus of the study was to investigate the fatigue of dental cements; the load was doubled and used for dynamic loading experiments.

Load character for dynamic loading on the other hand was up to 450N. Experimentally, it had been reported that 400N was used by many investigators as described earlier in the literature review section. The use of 450N in these experiments was selected to increase the loading as the pilot studies showed little effects can be happened to the luting cement under the used loading magnitude ascribed in the literature. Additionally, (Huysmans *et al* 1992) reported that the maximum bite forces, such as may occur during clenching and bruxing, can reach levels of ca. 450N.

In this study a cyclic rate of 2Hz was used. Chewing rates differ from one person to another and there is also a variation in the chewing rate within the same person. The rate is also dependent on the type of food. It was shown that fatigue characteristics of zinc phosphate depend on frequency; time to fracture which was shorter at a loading frequency of 20Hz compared to 1Hz (Okazaki and Nishimura, 1990).

Patterson and Johns (1992) reported that chewing takes place for a total of 20 minutes a day, therefore at a frequency of 2Hz, 2400 cycles per day could be expected. Therefore, 300,000 cycles simulated the number of chewing cycles in about 5 months.

8.1.5. Strain gauges as a measurement device:

Caputo and Standlee (1987) reported that electrical strain gauges are a more commonly used technique for strain measurements. They described their working principle as being based on their electrical resistance change when subjected to strain. Tension produces an increase in resistance, whilst compression causes a decrease. They added that, if such a strain gauge was bonded to the surface of a structure under load, monitoring the resistance change would produce knowledge of the strain at that point. The advantage of strain gauge use is a very accurate strain measurement capacity on the actual structure in simulated or real function. The drawbacks of strain gauges however, lie in the elaborate electrical equipment required to collect and record the strain data and more importantly that they are devices that measure strains at discrete points on the surface to which they are bonded. No direct strain information is obtained.

In these series of studies two strain gauges were installed on each gold crown. The location was about 1mm above the crown margin. It would have been better if there were more strain gauges on each crown. Two gauges were used because of the dimensions of the gauges and the available area on the axial surface of the gold crown. In addition to that, every strain gauge needed a soldering terminal also on the axial surface of the gold crown. The soldering terminal could be installed on the die for each strain gauge; however, before removing each gold crown from its die, the leads had to be cut. This would have then required re-installation of the gauges and that in turn might have affected the measurements. The strain gauges were not removed after their installation.

During some dynamic experiments under wet conditions, isolating material (Silicone adhesive) was applied to the strain gauges using a brush and to their connection wires on the soldering terminals. The isolating material was left for 48 hours to cure according to the manufacturer's recommendations. It was used for isolating corrosion sensitive electronic and electrical parts, involving the use of brass or copper from the liquid media with the aim of preventing liquid contamination and/or contact that might have affected the conductivity of the strain gauges.

The gold crowns on the higher convergence angle dies (24° TOC, 6mm axial height) had greater horizontal curvature of their axial walls. The strain gauges were designed to record strain on a flat surface and this produced some minor difficulties on handling them. Before starting the experiments, the offset of the amplifiers were zeroed. On some occasions, the amplifiers did not come to zero due to the curvature of the axial surfaces. In these instances the initial reading was recorded as zero micro-volts. This figure was subtracted from each measurement to give the value for strain when the crowns were loaded.

The amplified strain readings for static loading experiments were recorded manually for each run then put in Excel file format. The strain readings were adjusted for the amplifiers' calibration factors then plotted against the load.

8.1.6. Media selection for dynamic loading experiments:

The results from the static loading experiments had indicated little deterioration in the bond between the crowns and the dies. This was also true for the dynamic testing of the dry specimens. It was therefore decided to add aqueous media to the experimental variables using at first water and then acid.

For the dynamic loading experiments, a variety of aqueous media was used for testing. The mediums started from dry to wet experiments. The acid used was lactic acid with a pH of 2.74. This was chosen as an acid found intra-orally representing the bacterial products. Experimentally, this selection was based on Nomoto *et al* (2003) recommendations; it was lactic acid with sodium lactate buffer.

8.2. Further interpretations of the results:

8.2.1. The relation between total occlusal convergence and axial height:

There was an interesting finding that the increasing of the axial wall height by 2mm whilst maintaining a TOC of 24° affected the resultant strain on loading more sensitive than doubling the angle from 12° to 24°. Doubling the TOC did not increase the strain as much as changing the axial wall height only by 2mm. This finding emphasised the importance of occluso-cervical dimension compared with TOC during tooth preparation. However many papers have supported the importance of a low TOC, but few have been written regarding the occluso-cervial dimension. This work supported indirectly the earlier work of Smyd (1944) Kaufman *et al* (1961) and provided support for the clinical recommendation of Goodacre (2004).

In the static loading experiments, the marked decrease in the axial strain with the increased axial wall height within each different cementation condition compared to doubling the angle of taper; this might have been related to the surface area of the die. Doubling the angle of taper decreased the surface area only by (168.20/177.58) 94.72% whilst increasing the axial wall height increased the surface area by (213.34/168.20) 126.84%.

8.2.2. The relation between axial height and occlusal thickness of the casting:

The interesting point was the association between the amount of the occlusal reduction and the remaining axial wall height. In reducing a tooth more occlusally to make a casting thicker, this would be at the expense of axial wall height. The shorter preparation would increase the axial strain substantially.

8.3. Statistical analysis of the results:

Two strain gauges were used for each crown. To investigate the relation between each strain gauge reading to the complete build (sample) as a whole, an ANOVA test (Single Factor) was used to compare the slope of each run. The results showed the p values >0.05, which were not significant. The hypothetical question was whether the variance between the left and right strain gauges in each build (sample) was the same, less or more than the variance between builds (samples). At the 5% significant level, the expected slope of the strain gauge was the same for all the builds (samples), apart from some partially cemented samples: in such cases, this might have been related to the possible spread of cement onto the occlusal surface. In Section (7.1.1.1.1), 40% of the partially cemented crowns showed some spread of cement beyond the axio-occlusal line angle onto the occlusal surface.

The results of the ANOVA showed that there was greater variability in the build of an individual crown compared with variability between different builds. The tables of the statistical test analysis results were recorded in Section 12.2.

Other reasons might have been related to variations in the thickness of the luting cement and/or the gold crowns, skewing of the gold crown during seating at the time of the cementation, or a degree of non-axial loading during the experiments.

The conclusion was drawn that the strain gauges on each side of the crowns behaved in the same way.

8.4. Mathematical hypothesis for the experimental data:

The gold crown on its nickel chromium die roughly represented a hollow cup-shaped cylinder over a solid cylinder, although the shape was more precisely described as a frustum because of the taper of the side-walls. The hollow cap was held on the solid die using zinc phosphate cement; the bond relied on mechanical interlocking into the irregularities of the metal surfaces.

Loading was carried out with the crowns uncemented, partially cemented and fully cemented. The uncemented crowns had no cement, whilst the partially cemented had luting cement applied to the axial surfaces of the dies before the gold crown was seated. Lastly, for the fully cemented crowns the cement half filled the fitting surface of the crown and was then seated over the die.

The type of cementation affected the connection between the gold crown and its die. The uncemented assemblies represented no connection between the die and the gold crown. The connection rose with the increase in luting cement from partially to fully cemented. The partially cemented assemblies would have behaved as a sliding adhesive joint, whilst in the fully cemented, the crown-die assemblies were considered as one unit of solid composite cylinder (stiff joint).

The stiffness of the gold crown on its die was represented by the amount of applied load (Newton) against its given microstrain (N/µ). From the measurements of strain distribution against the applied loading with different cement cover (Section 7.1.1.1.2), the data showed that, at 220N the stiffness of the uncemented samples was 220/270.5215= (0.8132) N/µ; for the partially cemented 220/119.13081= (1.8467) N/µ and for the fully cemented samples was 220/7.4735113= (29.4373) N/µ.

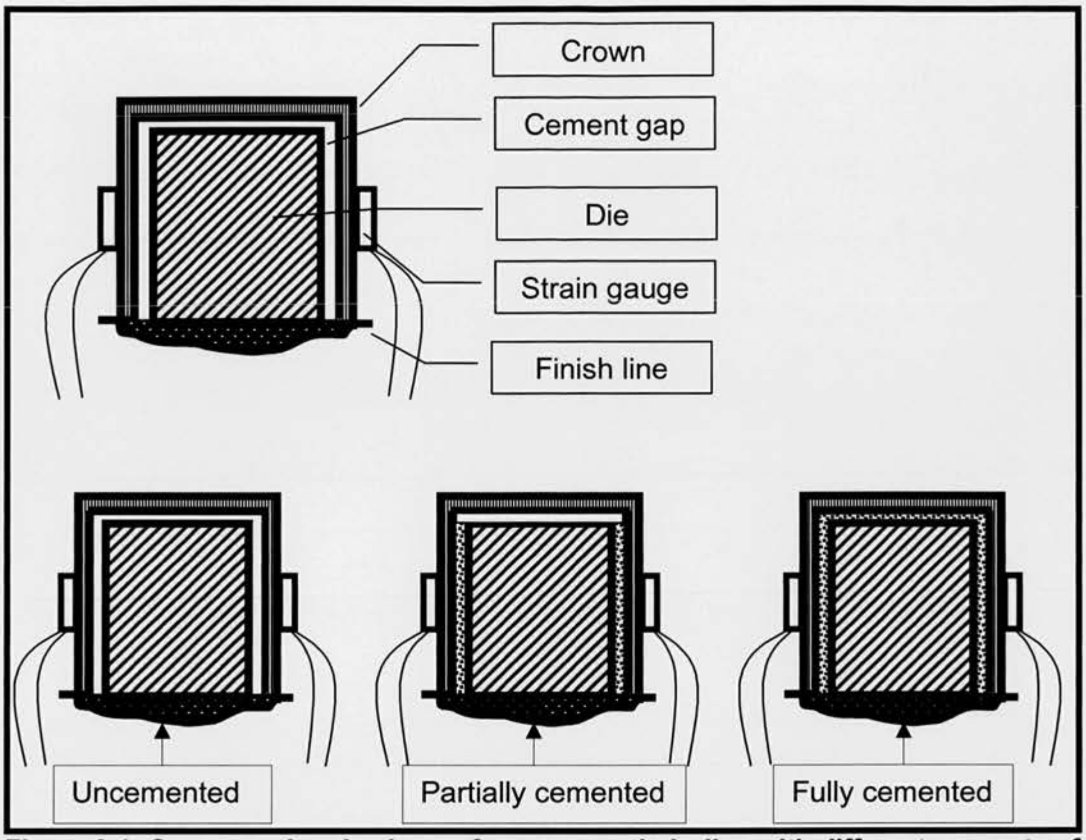


Figure 8.1: Cross sectionals views of crowns on their dies with different amounts of cement distribution.

A simple model of the stress distribution in the crown was developed as shown in Figure 8.1, where the crown was represented by an inverted cup shape sitting on the finish line. The effect of full cementation was to make the assembly behave as a solid composite cylinder, whereas the partially cemented crown would have behaved more like an adhesive joint.

In the uncemented case, the compressive stress would have been given approximately by the force divided by the annular cross-sectional area of the crown, giving an axial compressive strain of:

$$\varepsilon_u = \frac{F}{AE} = \frac{220}{\pi dtE} = \frac{220}{\pi \times 10 \times 0.5 \times 90 \times 10^3} \cong 160\mu\varepsilon$$

During application of axially directed load on the uncemented samples, since there was no luting cement between the crown and die, resistance to the load was provided by the die itself. The load concentration would have been at the crown margins where it articulated directly with the finish line of the die. The literature has recorded the inverse relationship between the amount of cement lute and the crown seating on its die (tooth) (Tan and Ibbetson, 1996).

For the partially-cemented crown, there was a shear strain on the luting cement on the axial surfaces of the crown-die assembly. A simplified solution for the tension in a single lap joint (Her S-C, 1999) was therefore used:

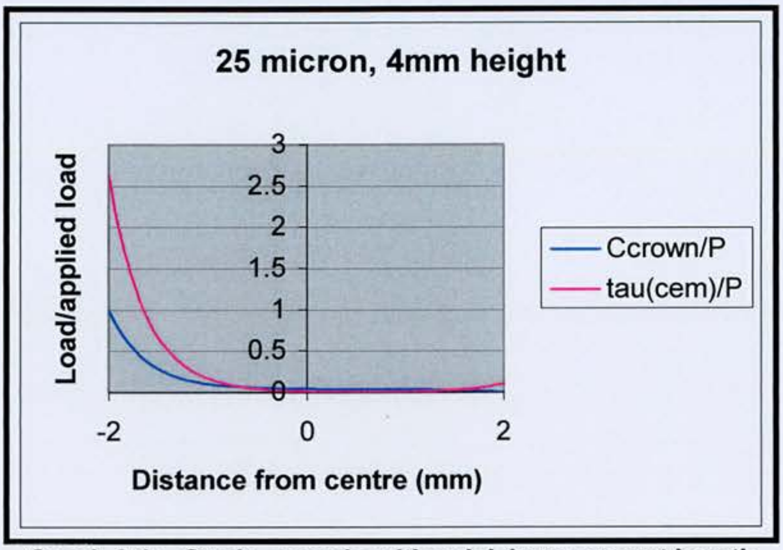
$$C_{crown} = P\left[-\frac{1}{2}\frac{\sinh(\lambda x)}{\sinh(\lambda l)} + \frac{E_{die}r_{die} - E_{crown}t_{crown}}{2(E_{die}r_{die} + E_{crown}t_{crown})}\frac{\cosh(\lambda x)}{\cosh(\lambda l)} + \frac{E_{crown}t_{crown}}{(E_{die}r_{die} + E_{crown}t_{crown})}\right]$$
$$= P\left[-\frac{1}{2}\frac{\sinh(\lambda x)}{\sinh(\lambda l)} + \frac{1}{2}C_1\frac{\cosh(\lambda x)}{\cosh(\lambda l)} + C_2\right]$$

$$\tau_{cement} = \frac{P\lambda}{2}\left[\frac{\cosh(\lambda x)}{\sinh(\lambda l)} - \frac{E_{die}r_{die} - E_{crown}t_{crown}}{(E_{die}r_{die} + E_{crown}t_{crown})}\frac{\sinh(\lambda x)}{\cosh(\lambda l)}\right]$$
$$= \frac{P\lambda}{2}\left[\frac{\cosh(\lambda x)}{\sinh(\lambda l)} - C_1\frac{\sinh(\lambda x)}{\cosh(\lambda l)}\right]$$

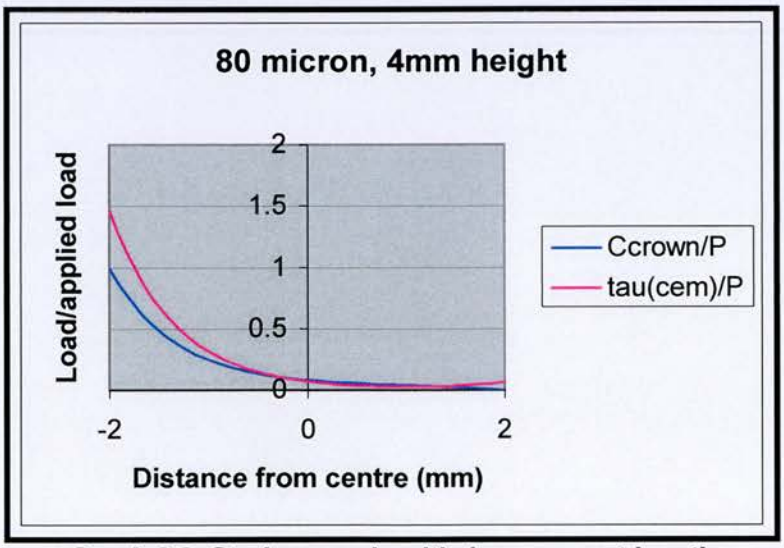
where the dimension *x* was measured axially from the approximate position of the strain gauge (mid-way axially from top to bottom of the cylinder, negative towards the top of the crown) and the parameter λ:

$$\lambda = \sqrt{\frac{G_{cem}}{t_{cem}}\left(\frac{1}{E_{die}r_{die}} + \frac{1}{E_{crown}t_{crown}}\right)} = t_{cem}^{-\frac{1}{2}}\sqrt{8.14\left(\frac{1}{220\times 5} + \frac{1}{90\times 0.5}\right)} = 0.434t_{cem}^{-\frac{1}{2}}mm^{-1}$$

Using E_{crown}=90GPa, E_{die}=220GPa, G_{cem}= 8.14GPa, t_{cem} = 25, 40, 80 micron, t_{crown}=0.5mm, t_{die}= 5mm and l=4mm, the following distribution of force in the crown and shear stress in the cement were calculated:



Graph 8.1: Strain records with axial 4mm cement length (25µ cement thickness)



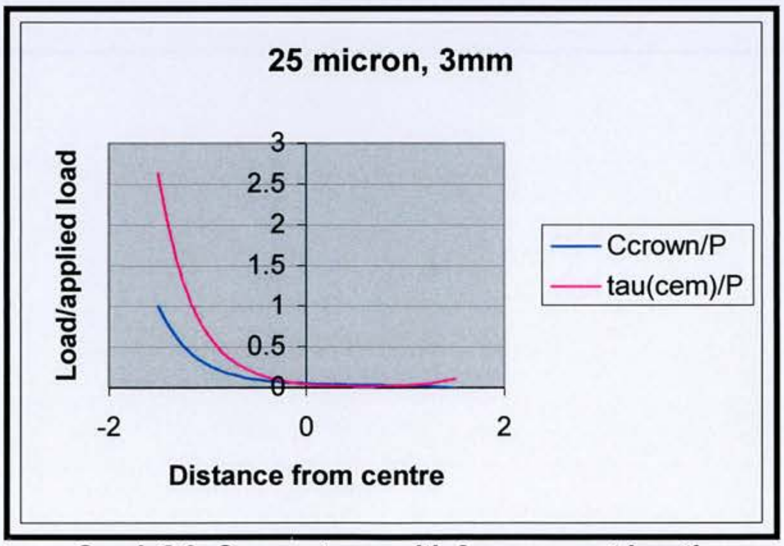
Graph 8.2: Strain records with 4mm cement length (80µ cement thickness)

In the cases of partially cemented and fully cemented samples, the strain concentration would have been at the two axial ends of the gold crown (edge of the occlusal surface and the crown margins) away from the central location of the strain gauges; the strain would have been concentrated on the occlusal surface of the crown and the crown margins where it articulated with the slope of the die's finish line.

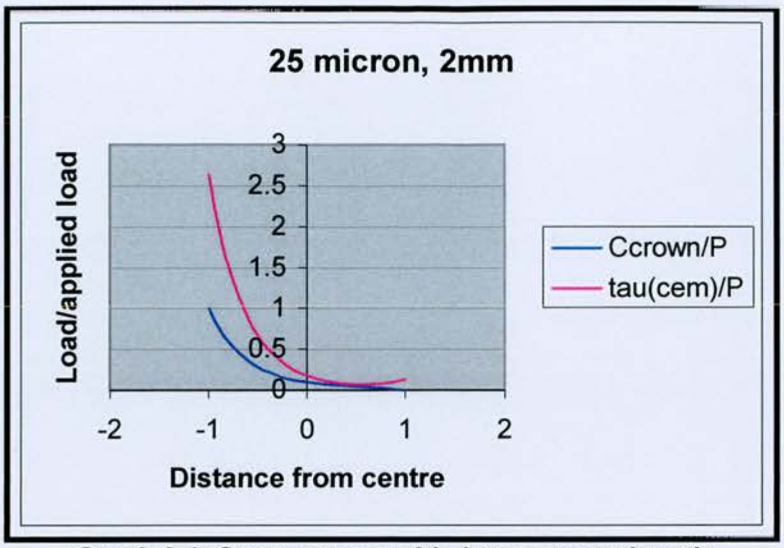
Zinc phosphate cement is water soluble, additionally, its solubility increases in acids of low pH. As a result, for the partially cemented samples, there were two gaps between the crown and its die. The first one was at the occlusal surface as the luting cement did not extend there during the cementation process; the second was at the margins, as zinc phosphate cement was soluble in wet media and was likely to have been lost from this area. Comparatively, the top gap has much greater effect on the resultant strain than the marginal one.

The position of the strain gauge would have influenced the recording of the strain in the axial wall as shown in Graphs 8.1, 8.2 and 8.4. The strain gauge was considered to be X mm from the margin and Y mm from the occlusal surface. C_{crown}/P (the ratio of actual load in the crown to the applied load) falls from unity to zero with a rapidity that depends on the degree of load transfer to the die, the thinner the cement layer the more efficient the load transfer Graphs 8.1 and 8.2. Since the measured compressive strain responds to C_{crown}, it would have been expected that thinner cement layers would have resulted in more efficient load transfer and hence lower measured strains for a given load. Generally, it would have been expected that a partially-cemented crown would have about half or less than half of the strain of an uncemented crown.

As can also be seen, the shear stress in the cement was maximum at the top of the crown and this will therefore be the focus of any damage under fatigue loading. If the cement deteriorated, the effective length of the joint will become shorter from the top and, as can be seen below in Graphs 8.1, 8.3 and 8.4, the crown stress will still fall to zero over the (shortened) cemented length; the part that has become uncemented will have a C_{crown}/P ratio of unity. Since the strain gauge remained in the same place, cement deterioration from the top of the crown could therefore be expected to increase the strain measured under a given load towards that expected for the uncemented crown.



Graph 8.3: Crown stress with 3mm cement length



Graph 8.4: Crown stress with 4mm cement length

For the fully cemented assemblies however, the types of strain would have been a combination of shear strain on the axial surfaces and compressive strain on the occlusal surface. As luting cements have a higher compressive compared to shear strength, there would have been less shear strain on the cement present on the axial surfaces. In other words, the cement that covered the occlusal surface had an essential role in the integrity of the cement film under a crown. For that reason the results revealed that the microstrain for the fully cemented samples was far lower than that of their uncemented and partially cemented counterparts.

A fully cemented crown could be regarded as a compound cylinder made up of three series elements I, II and III, and a compressive force, *F*, will be carried through all of the components as shown in Figure 8.2:

$F = F_I = F_{II} = F_{III}$

which will result in a change in length which is distributed between the elements as follows:

$$\delta = \delta_I + \delta_{II} + \delta_{III}$$

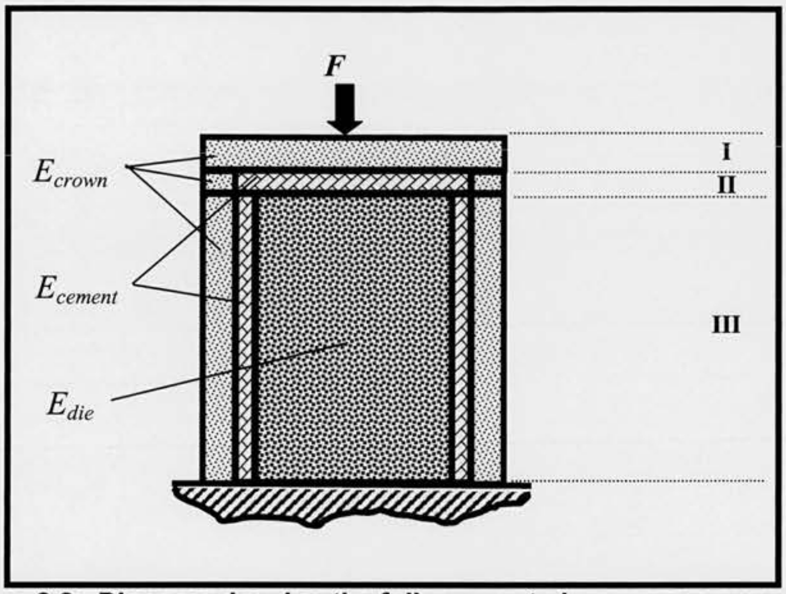


Figure 8.2: Diagram showing the fully cemented crown as compound structure under axial loading

Within the element II, the parallel sub-elements share the force, but have equal changes in length, so:

$$F_{II} = F_{crown} + F_{cement}$$
$$\delta_{II} = \delta_{crown} = \delta_{cement}$$

Also:

$$F_{crown} = E_{crown} A_{crown} \frac{\delta_{II}}{l_{II}}$$
$$F_{cement} = E_{cement} A_{cement} \frac{\delta_{II}}{l_{II}}$$

Therefore:

$$F_{II} = \frac{\delta_{II}}{l_{II}} \left[E_{crown} A_{crown} + E_{cement} A_{cement} \right]$$

and so the compressive strain in the crown can be expressed in terms of the compressive force, F:

$$\varepsilon_{crown} = \varepsilon_{II} = \frac{F}{E_{crown}A_{crown} + E_{cement}A_{cement}}$$

Within the element III, the parallel sub-elements share the force, but have equal changes in length, so:

$$F_{II} = F_{crown} + F_{cement} + F_{die}$$
$$\delta_{II} = \delta_{crown} = \delta_{cement} = \delta_{die}$$

Also:

$$F_{crown} = E_{crown}A_{crown}\frac{\delta_{II}}{l_{II}}$$
$$F_{cement} = E_{cement}A_{cement}\frac{\delta_{II}}{l_{II}}$$
$$F_{die} = E_{die}A_{die}\frac{\delta_{II}}{l_{II}}$$

Therefore:

$$F_{II} = \frac{\delta_{II}}{l_{II}}\left[E_{crown}A_{crown} + E_{cement}A_{cement} + E_{die}A_{die}\right]$$

and so the compressive strain in the crown can be expressed in terms of the compressive force, *F*:

$$\varepsilon_{crown} = \varepsilon_{III} = \frac{F}{E_{crown}A_{crown} + E_{cement}A_{cement} + E_{die}A_{die}}$$

$$\varepsilon_{fc} = \frac{220}{\pi dt_{crown}E_{crown} + \pi dt_{cement}E_{cement} + \pi d^2 E_{die}/4}$$
$$= \frac{220}{\pi \times 10 \times \left[0.5 \times 90 \times 10^3 + t_{cement} \times 20 \times 10^3 + 10 \times 220 \times 10^3\right]} \cong 3\mu\varepsilon$$

so the strain in a fully cemented crown is much lower than either the partially cemented or uncemented crown. Also, neither the cement nor the crown has much effect on the strain measured in the crown.

Comparing the calculated microstrain (160μ) for the uncemented crowns and (3μ) for the fully cemented crowns with the static experiment values in Graph 4.7, the calculated values were about the half of the experimental values 270.5μ for uncemented and 7.5μ for fully cemented crowns. This comparison was reasonable given the approximate nature of the calculation. Most importantly, the simple model predicted quite accurately the order of the effect between the uncemented and fully cemented crowns. This clarified the importance of the luting cement in load transfer.

The effect of taper is threefold in introducing a component of force directed at increasing contact between the crown and die, decreasing the force directed at compressing the crown wall, and in decreasing the cross-sectional area over the nominal diameter, d. For a taper of 6°, the diameter at the top of the crown is 8.74mm (0.874 of original xsa), and, for a taper of 12°, it is 7.45mm (0.745 of original xsa). The effect of the spread of the load is to produce a reduction in force of cosθ, 0.994 or 0.978 depending on the taper. The contact force component (sinθ) might be expected to increase load transfer and also contribute to a reduction in strain for a given load. It is clear, however, that the latter two effects are rather smaller than the effect on cross-sectional area, so the net effect of increased taper angle is expected to be an increase in the strain in the crown for a given load and given degree of cementation as shown in Figure 8.3.

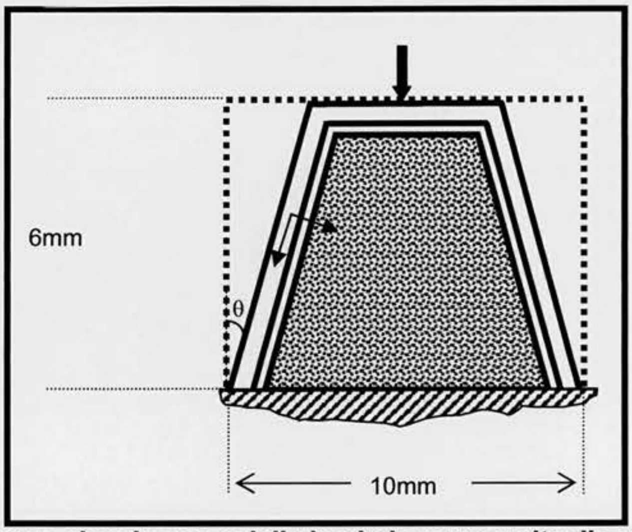


Figure 8.3: Diagram showing an axially loaded crown on its die and the effect of increased taper.

Mathematically, the relation between the compressive strains would be a matter of the cross-sectional area.

For the uncemented crown: $A = \pi rt$

For the fully cemented crown: $A = \pi r^2$

Using the average radius for each taper, which were 4.69mm for 6°, and 4.36mm for 12°.

For the uncemented crowns, the expected strain at any given load is 4.36/4.69 as much for the 6° taper. Comparing the results of Graph 7.7, the compressive strain at 220N for 6° taper was about 270 and for 12° about 340.

For fully cemented crowns, the expected strain at any given load is $4.36^2/4.69^2$ as much for the 6° taper. Comparing the results of Graph 7.9, the compressive strain at 220N was about 7.5 microstrain for 6° taper and about 13 microstrain for 12°. The compressive strain results obtained experimentally for both uncemented and fully cemented were not far away from the calculated ratios.

The hypothesis for the dynamic loading experiments was based on the luting cement bond degradation, with the stiffness decreasing from fully cemented toward partially and then uncemented.

There were multiple variables that influenced the results of testing the partially cemented samples:

1. Symmetry of the loading:

   The load was to be applied axially in the long axis of the crown-die assembly and the testing rig was designed in this way. However, the construction of the crowns might have allowed differences in their shape, adaptation and thickness. These factors could have resulted in crowns which were not identical.

2. Strain gauge Installation:

During strain gauge installation, the locations were opposite each other about 1mm above the crown margins perpendicular to the long axis of the crown-die assembly. The location was selected without specific measurements. This could have resulted in placements that were not identical between samples, which in turn might have affected the data recorded.

3. Quality of bonding:

There were many factors that could have affected the quality of the bond between the crown and die such as the thickness of the cement film, luting cement particle size, void distribution in the cement film and seating of the crown on its die.

4. Packing:

Possible changes in the behaviour of the cement during testing were a further variable. The phenomenon of "packing" of the cement was discussed in the literature review. If this occurred, the result would have been increased cement stiffness leading to a decrease in the compliance with time. This possibility had some support from the data for the microstrain measurements and the photomicrographs of the cement film. This possibility was most evident in the dynamic tests using purely acid media.

8.5. Limitations of the study:

1) The investigation was carried out *in vitro*.
2) The loads used during the dynamic tests were well in excess of those normally encountered intra-orally.
3) The design of the experiments did not explore the nature of the cement bond despite the sensitivity of the measuring strain gauges and the scanning electron microscope.
4) The experiments failed to show complete failure of any cemented crowns.

8.6. Clinical interpretation:

The function of the cement lute is to seal the gap between the fitting surface of the casting and the tooth preparation. In the case of the recurrent loading, deterioration of the cement at the margin would compromise the sealing effect; as a result microbial and ionic penetration will occur increasing the risks of dental caries and pulpal damage.

Under dynamic loading and in the presence of acidic media, the results indicated that the behaviour of the cement film did not remain consistent. There was a tendency for the cement to become less effective as evidenced by the sharp changes in microstrain. There was support for this from the electron micrographs although no direct correlation could be made.

The conditions in this series of experiments were not clinically representative, but the results indicated a hypothesis as to how the cement film behaves beneath a crown. The clinical relevance of this work is limited and this was expected. The significance of this research has been the establishment of a method by which the condition of the cement film beneath a crown can be monitored. It has demonstrated clearly the behaviour of traditional acid-base luting cement when subjected to loads and the resultant axial strains measured.

The continuation of this work would see the experimental model using axially located strain gauges being employed to examine the behaviour of different luting cements. If this were combined with different methods of loading restorations, further valuable data on the behaviour of cemented crowns would be obtained.

8.7. Recommended further work:

1. Using the same testing protocol on natural teeth or at least bovine dentine instead of metallic dies substrates.
2. Using the same methodology to investigate the effects of crown made from different materials.
3. Using the same methodology to compare the behaviour of different types of luting cements.
4. Work needs to be undertaken to investigate further the effects of cement distribution beneath crowns.

5. There would be benefit in using the same methodology but with non-axial loading.

Chapter 9: Conclusions

With the limitations of the experiments, the following conclusions could be made:

1. A standardised method was established which allowed fabrication of metal dies and cast gold crowns.
2. The method allowed reasonable control of the distribution of the cement between the die and the crown.
3. The use of strain gauges placed on the external surfaces of the gold crowns allowed the response of the crown-die assembly to axial loads to be recorded.
4. The method permitted the strain response of the axial walls of the crowns to be recorded when the crowns were fully cemented, partially cemented or uncemented.
5. Under static loading, the creation of increased space for the cement between the die and the fitting surface of fully cemented crowns resulted in increased compressive axial microstrain.
6. Under static loading, increasing the total occlusal convergence (taper) of the axial walls of the die from 12° to 24° resulted in a minor increase in the compressive axial microstrain.
7. When the taper of the die was 24°, increasing the axial wall height by 2mm resulted in a large decrease in compressive axial microstrain in the walls of the fully cemented crowns when tested under static loads.

8. Under static loading using a die of 6mm height and 12° taper, increasing the thickness of the occlusal surface of the gold crown from 0.5mm to 1.5mm increased the axial compressive strain in the crown by a factor of 2. Heat treatment of a crown of 0.5mm occlusal thickness tested under the same conditions did not alter the axial compressive micorstrain compared with the control.
9. The use of axially placed strain gauges on the external surfaces of the gold crowns allowed the opportunity to record axial microstrain for partially crowns tested using dynamic loading when the crown-die assemblies were tested in different media.
10. Dynamic testing of cemented samples stored dry had little effect on axial microstrain on one side of the crown and more on the other. The changes were increases in axial microstrain.
11. Hydration of the cement changed the strain response of the partially cemented crowns to dynamic loading.
12. The introduction of water as the medium for dynamic testing produced a decrease in the axial microstrain.
13. The use of acid and water and the acid alone during the dynamic tests altered the axial microstrain recorded as compared with crowns tested in air or water.
14. The microstrain results for the cemented crowns tested in acid showed some characteristic of those obtained for the uncemented samples in the static loading experiments.

15. In none of the dynamic tests was evidence obtained that the bond between the crown and the die had failed completely.
16. The experimental method demonstrated that the use of axially placed strain gauges on the external surfaces of a cast gold crown was a valid method for monitoring non-destructively the condition of the bond provided by the cement.

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Chapter 12: Appendix

12.1. Raw data for different experiments

12.1.1. Data of 12° total occlusal convergence and 6mm axial height

Uncemented samples

Load UC	Mean S1	Mean S2	Mean S3	Mean S4	Total mean
10	-6.310431	-8.932204	-10.56918	-5.88348	-7.9238
20	-11.91752	-24.262	-18.46052	-9.82242	-16.1156
30	-18.49688	-49.80995	-25.47435	-13.56467	-26.8364
40	-27.08929	-75.42268	-34.62191	-17.49988	-38.6584
50	-34.5202	-100.9371	-40.25877	-22.41855	-49.5337
60	-43.1238	-124.1471	-49.61795	-28.1389	-61.257
70	-50.74767	-146.2902	-56.50357	-33.6663	-71.8019
80	-58.867	-164.2567	-65.93124	-37.25288	-81.5769
90	-67.56894	-183.8266	-72.32511	-43.24217	-91.7407
100	-78.04857	-200.4386	-80.98093	-49.39831	-102.2166
110	-88.83443	-217.1825	-87.74949	-55.95529	-112.430
120	-102.7226	-233.3974	-96.77255	-63.48081	-124.0933
130	-117.4549	-250.5049	-103.1217	-70.93037	-135.5023
140	-132.6714	-265.191	-111.0369	-81.03526	-147.483
150	-160.6281	-278.0496	-120.41	-92.11989	-162.802
160	-168.1438	-287.4088	-133.6534	-103.704	-173.227
170	-188.7782	-297.6493	-149.1715	-116.2598	-187.963
180	-206.6314	-307.0047	-167.8478	-140.1773	-205.4153
190	-226.006	-316.4622	-188.9916	-155.3714	-221.7078
200	-244.5364	-325.1218	-209.4046	-174.044	-238.2767
210	-253.5958	-336.1416	-230.2757	-192.8	-253.2033
220	-279.6341	-347.6993	-245.5197	-209.233	-270.5215

Partially cemented samples

Load PC	Mean S1	Mean S2	Mean S3	Mean S4	Total mean
10	-5.928237	-2.70147	-2.720102	-2.894432	-3.56106
20	-11.36101	-5.013286	-5.644339	-6.3677	-7.0966
30	-17.27806	-7.041253	-8.769114	-10.2344	-10.83072
40	-22.7183	-8.982064	-13.44491	-14.7784	-14.9809
50	-28.71504	-11.20299	-17.92777	-21.1461	-19.748
60	-34.81013	-13.62807	-24.95278	-27.9832	-25.344
70	-40.70853	-15.37965	-31.4785	-35.6943	-30.8152
80	-46.21355	-17.22585	-38.01182	-43.9157	-36.3417
90	-52.11195	-18.78446	-46.58785	-51.3540	-42.2096
100	-58.19958	-20.5435	-53.49587	-58.6829	-47.7304
110	-64.17768	-22.49177	-62.65837	-66.6204	-53.9871
120	-70.56035	-24.05039	-71.90419	-73.1886	-59.926
130	-76.44383	-25.61273	-81.04428	-80.8311	-65.983
140	-83.012	-27.36431	-89.80594	-87.6869	-71.9673
150	-88.60417	-29.025	-98.74935	-95.8099	-78.047
160	-95.074	-30.68942	-105.158	-102.2797	-83.3003
170	-100.8629	-32.44473	-117.3432	-109.7330	-90.096
180	-107.8244	-34.21496	-124.4329	-116.5962	-95.767
190	-113.6096	-35.58434	-131.6172	-124.2462	-101.264
200	-119.9922	-37.35084	-139.5884	-131.2003	-107.033
210	-126.0799	-38.72022	-148.543	-139.3234	-113.167
220	-132.3716	-40.48299	-156.8203	-146.8489	-119.131

Fully cemented samples

Load FC	Mean S1	Mean S2	Mean S3	Mean S4	Total mean
10	-1.248659	-1.161503	-0.964795	-0.96481	-1.0849
20	-2.501048	-2.130043	-1.259832	-1.9259	-1.9542
30	-3.462129	-3.000237	-1.751562	-2.6940	-2.727
40	-4.143091	-3.386162	-1.751562	-2.7886	-3.0174
50	-4.816594	-3.870432	-2.042854	-3.3601	-3.5225
60	-5.104172	-3.968778	-2.1412	-3.4547	-3.667205
70	-5.679329	-4.45677	-2.432508	-4.4158	-4.2461
80	-5.679329	-4.555124	-2.432508	-4.5104	-4.294332
90	-6.451178	-4.941048	-2.920491	-4.8888	-4.8004
100	-6.545794	-4.941048	-2.920491	-4.8888	-4.824
110	-6.935448	-5.523664	-3.20807	-5.5549	-5.3055
120	-7.033794	-5.523664	-3.20807	-5.6495	-5.3538
130	-7.325102	-5.909588	-3.499378	-6.3193	-5.7633
140	-7.419718	-5.909588	-3.499378	-6.3193	-5.787
150	-7.900258	-6.495934	-3.69607	-6.989	-6.2703
160	-7.994875	-6.495934	-3.69607	-7.0837	-6.3176
170	-8.191567	-6.980204	-3.987461	-7.6551	-6.7036
180	-8.289912	-6.980204	-3.987461	-7.7497	-6.7518
190	-8.486604	-7.56655	-3.987461	-8.3211	-7.0904
200	-8.486604	-7.56655	-3.987461	-8.605	-7.1614
210	-8.872529	-7.759512	-4.085807	-9.1764	-7.4736
220	-8.872529	-7.759512	-4.085807	-9.1764	-7.4736

12.1.2. Data of stress-strain behaviour with different degrees of cement spacing

0 layers of die spacer

Load 0DS	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-1.0557	-0.768119	-2.879513	-0.976	-0.393384	-1.214543
20	-2.4064	-1.445351	-5.187599	-1.562346	-0.684691	-2.257277
30	-3.746	-1.918432	-7.102302	-2.152421	-1.078075	-3.199446
40	-4.9984	-2.013048	-8.827772	-2.447459	-1.172692	-3.891874
50	-5.8611	-2.588205	-9.883469	-2.742497	-1.464	-4.507854
60	-6.6292	-2.682821	-10.54951	-3.037534	-1.75158	-4.930129
70	-7.2006	-3.061286	-11.12467	-3.037534	-1.849924	-5.254803
80	-7.5828	-3.155902	-11.60521	-3.234226	-1.94827	-5.505282
90	-7.8629	-3.628983	-11.89279	-3.430918	-2.43627	-5.850372
100	-8.2489	-3.628983	-11.99113	-3.529264	-2.53089	-5.985833
110	-8.6273	-3.81822	-12.08575	-3.62761	-2.625502	-6.156876
120	-8.722	-3.912832	-12.38079	-3.922648	-2.91308	-6.37027
130	-9.1004	-4.196681	-12.4754	-3.922648	-3.109772	-6.560980
140	-9.4864	-4.291297	-12.77044	-4.11934	-3.499426	-6.833381
150	-9.6756	-4.575146	-12.9634	-4.11934	-3.597772	-6.986252
160	-9.7702	-4.858994	-13.25844	-4.316031	-3.794464	-7.199626
170	-9.8611	-4.760648	-13.35306	-4.316031	-3.89281	-7.23673
180	-10.05	-5.044497	-13.4514	-4.512723	-4.275005	-7.466725
190	-10.145	-5.139113	-13.54975	-4.611069	-4.570042	-7.602995
200	-10.24	-5.233729	-13.84106	-4.611069	-4.857621	-7.756696
210	-10.141	-5.612194	-13.9394	-4.709415	-4.955967	-7.871595
220	-10.236	-5.71054	-14.1361	-4.709415	-4.955967	-7.949604

10 layers of die spacer

Load 10DS	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-0.38593	-1.835005	-2.879513	-1.157773	-2.137502	-1.679145
20	-0.96482	-2.504778	-5.187599	-1.638313	-3.120961	-2.683294
30	-1.15778	-3.076205	-7.1023	-2.024238	-3.616421	-3.395389
40	-1.35075	-3.647632	-8.827772	-2.2172	-4.017264	-4.012124
50	-1.44909	-3.647632	-9.883469	-2.410161	-4.020994	-4.282269
60	-1.82754	-3.93521	-10.54951	-2.803545	-4.225145	-4.66819
70	-1.82754	-4.222789	-11.12467	-2.803545	-4.520183	-4.899745
80	-1.92588	-4.415751	-11.60521	-2.901891	-4.625988	-5.094944
90	-2.0205	-4.415751	-11.89279	-3.098583	-4.724334	-5.230392
100	-2.11512	-4.608713	-11.99113	-3.386162	-4.349599	-5.290145
110	-2.40643	-4.801675	-12.08575	-3.582853	-4.644636	-5.504269
120	-2.69401	-4.900021	-12.38079	-3.681199	-4.648366	-5.660877
130	-2.69401	-4.99837	-12.4754	-3.779545	-4.462863	-5.682038
140	-2.88698	-5.092983	-12.77044	-3.779545	-4.273631	-5.760716
150	-2.98532	-5.092983	-12.9634	-3.976237	-4.568669	-5.917322
160	-3.17829	-5.092983	-13.25844	-4.074583	-4.481512	-6.017162
170	-3.17829	-5.191329	-13.35306	-4.074583	-4.386896	-6.036832
180	-3.46212	-5.191329	-13.4514	-4.271275	-4.587317	-6.192688
190	-3.75342	-5.289675	-13.54975	-4.464237	-4.685663	-6.348549
200	-3.85177	-5.388021	-13.84106	-4.660929	-4.401814	-6.428719
210	-3.85177	-5.482637	-13.9394	-4.660929	-4.307198	-6.448387
220	-4.14308	-5.580983	-14.1361	-4.759275	-4.405544	-6.604996

20 layers of die spacer

Load 20DS	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-0.870202	-2.883243	-1.634584	-1.634584	-0.97227	-1.598975
20	-1.543696	-5.773945	-2.872054	-2.599394	-1.4603	-2.849878
30	-1.827542	-8.093221	-3.825675	-3.571664	-1.558616	-3.775344
40	-1.922162	-9.731534	-4.491719	-4.445588	-1.751578	-4.468516
50	-2.304351	-10.41623	-5.165221	-4.740626	-1.846194	-4.894524
60	-2.304351	-10.9005	-5.263567	-5.228626	-1.94454	-5.128317
70	-2.686543	-11.29015	-5.547416	-5.720356	-1.94081	-5.437055
80	-2.879505	-11.29015	-5.736648	-6.015393	-1.94081	-5.572501
90	-2.974125	-11.48311	-5.925881	-6.212085	-2.039156	-5.726871
100	-3.068745	-11.48311	-6.118843	-6.601739	-1.94081	-5.842649
110	-3.360047	-11.48311	-6.406421	-6.700085	-1.740389	-5.938010
120	-3.742242	-11.67607	-6.879502	-7.191815	-1.933351	-6.284596
130	-3.836862	-11.87277	-6.781156	-7.290161	-2.031697	-6.362529
140	-4.029828	-12.06573	-7.068735	-7.683544	-1.933351	-6.556238
150	-4.222794	-12.16034	-7.360043	-7.880236	-2.027967	-6.730276
160	-4.32114	-12.16034	-7.454659	-8.080658	-1.925892	-6.788538
170	-4.510372	-12.25496	-7.549275	-8.179004	-2.020508	-6.902824
180	-4.510372	-12.25496	-7.643891	-8.477771	-2.020508	-6.981500
190	-4.604982	-12.45165	-7.643891	-8.576117	-1.922162	-7.039760
200	-4.699592	-12.45165	-7.738508	-8.871155	-1.922162	-7.136613
210	-4.888828	-12.45165	-7.833124	-9.067847	-2.016778	-7.251645
220	-4.888828	-12.54627	-7.92774	-9.169922	-2.111394	-7.328831

12.1.3. Data of 24° total occlusal convergence and 6mm axial height

Uncemented samples

Load UC	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-2.818457	-31.97027	-6.766237	-2.818464	-11.6164	-11.19797
20	-25.44072	-57.69588	-15.71238	-25.44077	-26.41108	-30.1402
30	-55.83367	-82.05583	-27.77948	-55.83378	-42.23162	-52.7469
40	-77.36667	-104.9406	-42.06751	-77.36683	-55.82377	-71.5131
50	-99.01667	-127.4245	-55.95843	-99.01686	-66.82771	-89.6488
60	-115.4543	-148.0972	-69.263	-115.4546	-74.62217	-104.578
70	-134.7739	-170.1392	-85.03368	-134.7741	-82.95448	-121.5351
80	-148.3246	-188.5275	-100.8841	-148.3249	-91.0901	-135.430
90	-165.0447	-209.6547	-116.6324	-165.0451	-99.41495	-151.158
100	-177.9033	-228.3082	-132.6682	-177.9036	-111.5431	-165.665
110	-193.1445	-247.741	-150.5205	-193.1449	-124.8066	-181.8715
120	-204.7283	-267.0606	-167.1166	-204.7287	-138.5545	-196.438
130	-220.2683	-285.5897	-182.1951	-220.2687	-154.3838	-212.541
140	-231.7612	-314.2484	-197.3981	-231.7617	-168.2337	-228.681
150	-246.817	-323.3139	-211.5491	-246.8174	-182.7459	-242.249
160	-256.445	-339.4901	-226.0064	-256.4455	-194.1283	-254.503
170	-272.2913	-360.2215	-239.8959	-272.2918	-209.507	-270.8415
180	-282.7135	-377.858	-253.6871	-282.7141	-202.2649	-279.848
190	-298.3519	-394.9878	-264.5727	-298.3524	-236.6764	-298.588
200	-308.8427	-409.3938	-276.6957	-308.8433	-247.2147	-310.198
210	-323.8001	-423.8683	-285.7575	-323.8007	-263.8719	-324.2197
220	-335.7661	-437.6656	-294.1532	-335.7667	-276.1394	-335.8982

Partially cemented

Load PC	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-3.269167	-5.111632	-3.957589	-5.265923	-5.410399	-4.602942
20	-6.065253	-10.12492	-7.922637	-8.878614	-8.929848	-8.38425
30	-8.952226	-15.78933	-12.17899	-12.58592	-16.17525	-13.1363
40	-13.00816	-21.63924	-16.81754	-16.68661	-21.83966	-17.9982
50	-17.06782	-28.85108	-21.55817	-20.59061	-28.91959	-23.3975
60	-21.32418	-36.05918	-26.49176	-24.6913	-33.65648	-28.44458
70	-26.54162	-41.90537	-31.02823	-29.76053	-46.32858	-35.1129
80	-31.28224	-47.57351	-36.83575	-35.1248	-52.21579	-40.6064
90	-37.48687	-53.91888	-41.86022	-39.411	-59.50359	-46.4361
100	-42.60223	-59.8746	-46.97931	-44.57857	-72.88884	-53.38471
110	-48.40228	-65.5502	-52.48806	-49.34904	-79.3251	-59.023
120	-54.0131	-70.93822	-57.99307	-54.61496	-93.37915	-66.1877
130	-59.90777	-76.23908	-63.99355	-59.87342	-102.7583	-72.5544
140	-65.51487	-80.86644	-69.11264	-65.92239	-108.2946	-77.94212
150	-71.41699	-85.29338	-75.01104	-71.67258	-117.8681	-84.2524
160	-77.20959	-89.44393	-80.1264	-77.03312	-125.1894	-89.8005
170	-83.00964	-94.37006	-86.31238	-92.52878	-133.1694	-97.8781
180	-88.70762	-98.42227	-91.61697	-98.46821	-140.2903	-103.5012
190	-94.60602	-102.7695	-96.83068	-104.2109	-149.7953	-109.643
200	-100.1185	-106.6362	-101.844	-110.1466	-157.3968	-115.228
210	-106.221	-111.0818	-108.7146	-116.279	-162.7513	-121.001
220	-111.6538	-115.3307	-114.1101	-121.9271	-169.4192	-126.4882

Fully cemented

Load FC	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
10	-1.536229	-1.15777	-2.020508	-1.536238	-1.528778	-1.55591
20	-2.686536	-2.87578	-3.946399	-2.686551	-2.762519	-2.991557
30	-3.836843	-4.89256	-6.549523	-3.836864	-3.803297	-4.58382
40	-4.794189	-7.08738	-8.286183	-4.794216	-5.040767	-6.000547
50	-5.84988	-9.19132	-10.40877	-5.849913	-5.805156	-7.42101
60	-6.71261	-10.7164	-11.56281	-6.712648	-6.089005	-8.3587
70	-7.189418	-11.6737	-12.91354	-7.189459	-6.755048	-9.144233
80	-7.764571	-12.9149	-13.68539	-7.764615	-7.038897	-9.83368
90	-8.143033	-13.7739	-14.55186	-8.14308	-7.417362	-10.4059
100	-8.335994	-14.4474	-15.03986	-8.336042	-7.610324	-10.7539
110	-8.623571	-14.8259	-15.90632	-8.623621	-7.70494	-11.13687
120	-8.816532	-15.3973	-16.10301	-8.816583	-7.799556	-11.3866
130	-9.104109	-15.5903	-16.6819	-9.104161	-7.988789	-11.6939
140	-9.29334	-15.7832	-17.06783	-9.293394	-8.272637	-11.94208
150	-9.48257	-16.0708	-17.45375	-9.482626	-8.367254	-12.1714
160	-9.671801	-16.2638	-17.74133	-9.671859	-8.367254	-12.34321
170	-9.770147	-16.453	-18.2256	-9.770205	-8.46187	-12.53616
180	-10.05399	-16.453	-18.51691	-10.05405	-8.560216	-12.72763
190	-10.14861	-16.5476	-18.80075	-10.14867	-8.749448	-12.879
200	-10.24322	-16.7369	-18.80075	-10.24329	-9.033297	-13.0115
210	-10.43619	-17.0244	-19.18295	-10.43625	-9.033297	-13.2226
220	-10.43619	-17.2174	-19.27756	-10.43625	-9.033297	-13.2801

12.1.4. Data of 24° total occlusal convergence and 8mm axial height

Uncemented samples

Load UC	Mean S1	Mean S2	Mean S 3	Mean S 4	Mean S5	Total mean
10	-34.98248	-3.605232	-8.553348	-10.18125	-4.744356	-12.41333
20	-66.29633	-7.815458	-20.45046	-24.50324	-18.77936	-27.56897
30	-76.8906	-10.8418	-28.58431	-34.48781	-29.43566	-36.04804
40	-87.91182	-14.36734	-40.1803	-42.54511	-42.75103	-45.55112
50	-93.49516	-17.10983	-56.80586	-49.95874	-53.23301	-54.12052
60	-105.3605	-19.65191	-74.79472	-56.7026	-64.20673	-64.14329
70	-112.7191	-22.1031	-96.12126	-63.44645	-73.22845	-73.52367
80	-123.9557	-25.03856	-116.9523	-70.67458	-81.89781	-83.70379
90	-131.4887	-27.87567	-128.1157	-78.69321	-60.54045	-85.3428
100	-141.4803	-30.6219	-156.6438	-86.89734	-92.97831	-101.724
110	-148.7332	-33.46274	-168.9761	-94.52631	-96.05177	-108.350
120	-156.9919	-36.60235	-197.2242	-101.5727	-99.57966	-118.394
130	-164.2634	-39.54154	-218.5022	-108.8083	-100.1073	-126.245
140	-173.1047	-43.06707	-236.0306	-125.0287	-102.2397	-135.89
150	-179.5969	-46.58888	-256.0562	-132.3738	-102.7263	-143.468
160	-187.1821	-50.40199	-271.2616	-130.9494	-104.5674	-148.8725
170	-194.1735	-54.50641	-291.5786	-140.5267	-115.1538	-159.1878
180	-202.2467	-58.41414	-308.9102	-147.9776	-106.8504	-164.88
190	-209.3402	-62.80614	-326.9042	-156.4768	-107.3184	-172.569
200	-218.1055	-67.09233	-342.4917	-166.0689	-109.7197	-180.696
210	-225.3882	-72.3396	-361.931	-173.0743	-110.9558	-188.738
220	-234.2407	-77.57942	-380.5113	-185.6928	-112.6873	-198.1423

Partially cemented samples

Load PC	Mean S1	Mean S2	Mean S 3	Mean S4	Mean S5	Total mean
10	-2.14123	-1.558616	-3.163362	-2.777438	-1.649502	-2.25803
20	-3.794464	-3.018886	-5.070605	-5.256108	-3.20066	-4.068145
30	-5.35308	-4.184118	-6.879502	-7.356313	-4.467967	-5.648196
40	-7.108388	-5.549772	-9.358172	-9.645751	-5.542312	-7.440879
50	-9.064117	-6.624117	-12.03353	-11.74596	-6.522042	-9.19795
60	-11.21281	-7.600117	-14.31924	-14.41013	-7.308809	-10.97022
70	-13.75488	-8.97323	-16.51033	-16.79418	-7.793079	-12.76514
80	-15.51765	-9.067847	-18.70143	-18.989	-8.962041	-14.2476
90	-18.15807	-10.04385	-21.27472	-21.94821	-9.544657	-16.1939
100	-20.70388	-10.72854	-23.6625	-23.94634	-10.51693	-17.91164
110	-22.56126	-11.60992	-25.37305	-26.14117	-11.0012	-19.33732
120	-25.2963	-12.58219	-27.46952	-28.33599	-11.87512	-21.1118
130	-27.25576	-13.56192	-29.55854	-30.43246	-12.55608	-22.673
140	-29.70321	-14.05365	-31.5604	-32.80906	-13.23332	-24.27193
150	-32.14321	-14.73834	-33.74403	-35.19684	-14.10351	-25.9852
160	-34.58694	-15.616	-36.69205	-37.67551	-14.78074	-27.87025
170	-36.93233	-16.3953	-38.49349	-40.15791	-15.55632	-29.50707
180	-39.66736	-16.98538	-40.77547	-42.63658	-16.3319	-31.27934
190	-41.82351	-17.76842	-42.48602	-45.11152	-16.71782	-32.78146
200	-44.16517	-18.64607	-45.23362	-47.87404	-17.20209	-34.6242
210	-46.31759	-19.42538	-46.56198	-50.92414	-17.78844	-36.20351
220	-48.56463	-20.01172	-48.2688	-53.20985	-18.18182	-37.6474

Fully cemented samples

Load FC	Mean S1	Mean S 2	Mean S 3	Mean S 4	Mean S5	Total mean
10	-1.154043	-1.358194	-0.870194	-0.862735	-0.976	-1.044233
20	-2.493589	-2.228389	-1.744119	-2.016778	-1.755308	-2.04764
30	-3.640173	-2.811005	-2.610583	-2.883243	-2.636691	-2.916339
40	-4.881373	-3.681199	-3.480778	-3.745978	-3.317653	-3.8214
50	-6.126302	-3.874161	-3.870432	-4.510367	-3.900269	-4.45631
60	-6.977848	-4.744356	-4.551394	-5.566064	-4.490345	-5.2660
70	-7.942659	-5.13401	-5.12655	-6.235837	-5.175037	-5.92282
80	-8.888821	-5.516204	-5.417859	-6.90188	-5.56842	-6.45864
90	-9.551134	-5.803783	-5.803783	-7.571653	-5.954345	-6.93694
100	-10.31179	-6.095091	-6.095091	-8.241426	-6.54069	-7.45682
110	-10.97784	-6.481015	-6.288053	-8.820313	-7.127036	-7.93885
120	-11.74223	-6.866939	-6.677707	-9.584702	-7.51669	-8.477654
130	-12.412	-7.059902	-6.677707	-9.970626	-7.807998	-8.78565
140	-12.9797	-7.635058	-6.677707	-10.34909	-8.201382	-9.16859
150	-13.55112	-7.635058	-6.969015	-11.11721	-8.594766	-9.573434
160	-14.2209	-8.119328	-7.642518	-11.40479	-8.886074	-10.0547
170	-14.60682	-8.308561	-7.740864	-11.78698	-9.370344	-10.3627
180	-14.98529	-8.599869	-7.740864	-12.27125	-9.46869	-10.6132
190	-15.36748	-8.891177	-7.83548	-12.64972	-9.854614	-10.9197
200	-15.93891	-9.269642	-8.032172	-13.03564	-9.95296	-11.24587
210	-16.12814	-9.367988	-8.418096	-13.51618	-10.14592	-11.51527
220	-16.41199	-9.659296	-8.705674	-13.90211	-10.24427	-11.78467

12.1.5. Heat treatment and increased occlusal thickness

Heat treated samples

Load Heat	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
20	-2.515967	-1.656962	-1.627124	-1.933351	-1.335816	-1.813844
40	-4.16547	-2.625502	-2.092746	-2.807275	-2.005589	-2.739316
60	-5.13401	-2.822194	-2.471211	-3.491967	-2.777438	-3.339364
80	-6.004204	-3.310194	-2.354216	-4.558853	-3.250519	-3.895597
100	-6.58682	-3.597772	-2.732681	-4.952237	-3.636443	-4.301191
120	-6.980204	-4.278734	-3.107416	-5.735275	-4.113254	-4.842977
140	-7.661166	-4.37708	-3.485881	-6.416237	-4.499178	-5.28791
160	-8.342128	-4.86508	-3.860616	-7.195544	-4.877643	-5.828202
180	-8.633436	-5.251004	-4.046119	-7.880236	-5.161491	-6.194457
200	-9.219782	-5.83362	-4.322508	-8.76162	-5.543686	-6.736243
220	-9.50736	-6.124928	-4.700973	-9.347966	-6.024227	-7.141091
240	-10.08998	-6.707545	-4.882746	-10.02893	-6.595654	-7.660971
260	-10.47963	-7.097199	-4.969903	-10.81196	-6.784886	-8.028716
280	-11.06598	-7.585198	-5.250022	-11.30369	-7.265426	-8.494063
300	-11.35728	-7.78189	-5.628487	-11.79542	-7.360043	-8.784624
320	-11.94363	-8.171544	-6.003222	-12.66562	-7.833124	-9.323428
340	-11.94363	-8.368236	-6.090379	-13.44866	-8.120702	-9.594321
360	-12.52998	-8.947123	-6.56346	-14.31885	-8.404551	-10.15279
380	-12.82128	-9.140085	-6.658076	-14.9052	-8.783016	-10.46153
400	-13.21467	-9.533468	-6.934465	-15.49527	-8.764367	-10.78845
420	-13.60805	-9.631814	-7.025352	-16.17623	-9.551134	-11.19852
440	-14.28901	-10.21443	-7.403817	-16.95181	-10.02422	-11.77666
460	-14.58032	-10.40739	-7.490973	-17.24685	-10.21718	-11.98854

Increased occlusal thickness

Load O.Thickness	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
20	-2.489859	-2.319275	-1.7292	-1.180151	-2.781167	-2.099930
40	-5.18387	-4.339783	-2.679092	-2.742497	-5.740378	-4.137124
60	-7.003956	-6.269404	-3.534367	-3.529264	-8.779286	-5.823255
80	-9.297123	-8.10441	-4.295027	-4.494075	-12.68839	-7.775805
100	-11.3139	-9.262182	-4.866454	-4.982074	-15.26167	-9.137256
120	-12.84641	-10.80588	-5.532497	-5.56842	-17.44904	-10.44045
140	-14.48472	-11.47192	-6.100194	-6.253112	-18.87574	-11.43714
160	-16.804	-12.91354	-6.667892	-7.02869	-20.30617	-12.74406
180	-18.05639	-13.96924	-7.140973	-7.422074	-21.25979	-13.56969
200	-19.89139	-15.12701	-7.614054	-8.00469	-22.1188	-14.55119
220	-20.95828	-15.60382	-8.087135	-8.299728	-22.59188	-15.10817
240	-22.21813	-16.46656	-8.654832	-8.98069	-23.45462	-15.95497
260	-23.37217	-17.1326	-9.324605	-9.275728	-23.83308	-16.58764
280	-24.43906	-17.99534	-9.896032	-10.05131	-24.59747	-17.39584
300	-25.11256	-18.47215	-10.36911	-10.34634	-24.97593	-17.85522
320	-26.17572	-19.14192	-10.84219	-10.93269	-25.44901	-18.50831
340	-26.95129	-19.72081	-11.41362	-11.42069	-25.73286	-19.04785
360	-27.62853	-20.48893	-11.98132	-12.10165	-26.50098	-19.74028
380	-28.59707	-20.96201	-12.4544	-12.4913	-26.87945	-20.27685
400	-29.46726	-21.73013	-12.73452	-13.07765	-27.35253	-20.87242
420	-30.04615	-22.01398	-12.92002	-13.27434	-27.73472	-21.19784
440	-30.82173	-22.39617	-13.48772	-13.9553	-28.30242	-21.79267
460	-31.306	-22.87298	-13.86245	-14.152	-28.49165	-22.13702

Control samples

Load Control	Mean S1	Mean S2	Mean S3	Mean S4	Mean S5	Total mean
20	-2.515967	-2.224659	-1.445351	-2.281978	-1.259848	-1.945561
40	-4.449318	-3.193199	-2.304356	-2.384054	-1.751578	-2.816501
60	-3.647632	-3.870432	-3.079935	-4.844075	-2.141232	-3.516661
80	-5.705437	-4.555123	-3.655091	-5.411773	-2.43254	-4.351993
100	-6.575631	-4.941048	-4.131902	-5.9832	-2.92054	-4.910464
120	-7.052442	-5.523664	-4.608713	-6.456281	-3.208118	-5.369844
140	-7.438366	-5.909588	-5.285945	-7.027708	-3.499426	-5.832207
160	-7.922637	-6.495934	-5.770215	-7.406173	-3.696118	-6.258215
180	-8.206485	-6.980204	-6.15241	-7.686292	-3.987426	-6.602563
200	-8.395718	-7.56655	-6.632951	-8.061027	-3.987426	-6.928734
220	-8.683296	-7.759512	-7.208107	-8.541567	-4.085772	-7.255651
240	-9.06922	-8.345858	-7.87788	-8.916302	-4.282464	-7.698345
260	-9.163837	-8.640896	-7.972496	-9.294767	-4.570042	-7.928408
280	-9.546031	-9.223512	-8.547653	-9.676962	-4.952237	-8.389279
300	-9.738993	-9.51482	-9.130269	-9.960811	-4.952237	-8.659426
320	-10.02657	-10.19205	-9.508734	-10.53597	-5.148929	-9.082451
340	-10.11746	-10.48709	-9.894658	-11.01651	-5.148929	-9.332929
360	-10.40504	-10.97882	-10.28058	-11.59166	-5.247275	-9.700675
380	-10.49965	-10.97882	-10.66278	-11.87924	-5.345621	-9.873222
400	-10.59427	-11.7544	-11.13959	-12.54902	-5.443966	-10.29625
420	-10.69261	-12.14405	-11.42344	-12.64363	-5.636929	-10.50813
440	-10.78723	-12.7304	-11.90025	-13.11671	-5.928237	-10.89257
460	-10.97646	-12.82874	-12.19156	-13.30967	-6.026583	-11.06660

12.2. Statistical analysis of static and dynamic data

ANOVA OF UNCEMENTED ASSEMBLIES FOR ALL GEOMETRIES

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1.12	2	2.6227	1.31135	0.195625
S2.12	2	3.2256	1.6128	0.895657
S3.12	2	2.0092	1.0046	1.15E-05
S4.12	2	1.8429	0.92145	0.0252
S1.24	2	3.0093	1.50465	1.5437
S2.24	2	3.8581	1.92905	3.156328
S3.24	2	2.9131	1.45655	0.145422
S4.24	2	3.0093	1.50465	1.5437
S5.24	2	2.4435	1.22175	0.598527
S1.8	2	1.7125	0.85625	0.085657
S2.8	2	0.6656	0.3328	0.115392
S3.8	2	3.7156	1.8578	0.550201
S4.8	2	1.5251	0.76255	0.550096
S5.8	2	1.2693	0.63465	0.6327

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	5.790015	13	0.445386	0.621166	**0.800933**	2.507263
Within Groups	10.03822	14	0.717016			
Total	15.82823	27				

ANOVA OF PARTIALLY CEMENTED ASSEMBLIES FOR ALL GEOMETRIES

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1.12	2	1.2085	0.60425	0.006647
S2.12	2	0.354	0.177	0.017447
S3.12	2	1.5522	0.7761	0.274689
S4.12	2	1.4235	0.71175	0.008965
S1.24	2	1.0725	0.53625	0.009207
S2.24	2	1.064	0.532	0.006774
S3.24	2	1.0642	0.5321	0.005682
S4.24	2	1.139	0.5695	0.102786
S5.24	2	1.6653	0.83265	0.03996
S1.8	2	0.4549	0.22745	0.050467
S2.8	2	0.1682	0.0841	0.006161
S3.8	2	3.7156	1.8578	0.550201
S4.8	2	0.4754	0.2377	0.04067
S5.8	2	0.1526	0.0763	0.001824

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	5.307221	13	0.408248	5.096373	**0.002346**	2.507263
Within Groups	1.121478	14	0.080106			
Total	6.428699	27				

ANOVA OF FULLY CEMENTED ASSEMBLIES FOR ALL GEOMETRIES

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1.12	2	0.0627	0.03135	4.81E-06
S2.12	2	0.0564	0.0282	0.000192
S3.12	2	0.0295	0.01475	9.38E-05
S4.12	2	0.0734	0.0367	0.000849
S1.24	2	0.0739	0.03695	0.000515
S2.24	2	0.1337	0.06685	0.001978
S3.24	2	0.145	0.0725	9.8E-07
S4.24	2	0.0739	0.03695	0.000515
S5.24	2	0.0579	0.02895	0.001017
S1.8	2	0.1421	0.07105	0.003741
S2.8	2	0.0737	0.03685	4.5E-08
S3.8	2	0.0648	0.0324	5.83E-05
S4.8	2	0.1201	0.06005	0.000853
S5.8	2	0.0872	0.0436	0.001383

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.008257	13	0.000635	0.793829	**0.658463**	2.507263
Within Groups	0.011202	14	0.0008			
Total	0.019459	27				

ANOVA for 12°&6mm axial height uncemented

S1	S2	S3	S4
0.9986	0.9436	1.007	0.8092
1.6241	2.282	1.0022	1.0337

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	2.6227	1.31135	0.195625
S2	2	3.2256	1.6128	0.895657
S3	2	2.0092	1.0046	1.15E-05
S4	2	1.8429	0.92145	0.0252

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.595888	3	0.198629	0.711618	0.593953	6.591382
Within Groups	1.116494	4	0.279124			
Total	1.712382	7				

ANOVA for 12°&6mm axial height partially cemented

S1	S2	S3	S4
0.5466	0.0836	0.4055	0.6448
0.6619	0.2704	1.1467	0.7787

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	1.2085	0.60425	0.006647
S2	2	0.354	0.177	0.017447
S3	2	1.5522	0.7761	0.274689
S4	2	1.4235	0.71175	0.008965

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.436325	3	0.145442	1.890404	0.272364	6.591382
Within Groups	0.307747	4	0.076937			
Total	0.744073	7				

ANOVA for 12°&6mm axial height fully cemented

S1	S2	S3	S4
0.0298	0.0184	0.0079	0.0573
0.0329	0.038	0.0216	0.0161

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	0.0627	0.03135	4.81E-06
S2	2	0.0564	0.0282	0.000192
S3	2	0.0295	0.01475	9.38E-05
S4	2	0.0734	0.0367	0.000849

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.000525	3	0.000175	0.613782	0.641226	6.591382
Within Groups	0.001139	4	0.000285			
Total	0.001664	7				

ANOVA for 24°&6mm axial height uncemented

S1	S2	S3	S4	S5
0.6261	0.6728	1.7262	0.6261	1.7688
2.3832	3.1853	1.1869	2.3832	0.6747

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	3.0093	1.50465	1.5437
S2	2	3.8581	1.92905	3.156328
S3	2	2.9131	1.45655	0.145422
S4	2	3.0093	1.50465	1.5437
S5	2	2.4435	1.22175	0.598527

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.521433	4	0.130358	0.093277	0.98035	5.192168
Within Groups	6.987678	5	1.397536			
Total	7.509112	9				

ANOVA for 24°&6mm axial height partially cemented

S1	S2	S3	S4	S5
0.4684	0.5902	0.4788	0.3428	0.974
0.6041	0.4738	0.5854	0.7962	0.6913

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	1.0725	0.53625	0.009207
S2	2	1.064	0.532	0.006774
S3	2	1.0642	0.5321	0.005682
S4	2	1.139	0.5695	0.102786
S5	2	1.6653	0.83265	0.03996

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.136707	4	0.034177	1.039382	0.470303	5.192168
Within Groups	0.164409	5	0.032882			
Total	0.301116	9				

ANOVA for 24°&6mm axial height fully cemented

S1	S2	S3	S4	S5
0.053	0.0983	0.0718	0.053	0.0515
0.0209	0.0354	0.0732	0.0209	0.0064

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	0.0739	0.03695	0.000515
S2	2	0.1337	0.06685	0.001978
S3	2	0.145	0.0725	9.8E-07
S4	2	0.0739	0.03695	0.000515
S5	2	0.0579	0.02895	0.001017

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.003123	4	0.000781	0.969622	0.497921	5.192168
Within Groups	0.004027	5	0.000805			
Total	0.00715	9				

ANOVA for 24°&8mm axial height uncemented

S1	S2	S3	S4	S5
0.6493	0.0926	1.3333	1.287	0.0722
1.0632	0.573	2.3823	0.2381	1.1971

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	1.7125	0.85625	0.085657
S2	2	0.6656	0.3328	0.115392
S3	2	3.7156	1.8578	0.550201
S4	2	1.5251	0.76255	0.550096
S5	2	1.2693	0.63465	0.6327

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	2.659376	4	0.664844	1.718791	0.281532	5.192168
Within Groups	1.934045	5	0.386809			
Total	4.59342	9				

ANOVA for 24°&8mm axial height partially cemented

S1	S2	S3	S4	S5
0.0686	0.0286	1.3333	0.3803	0.0461
0.3863	0.1396	2.3823	0.0951	0.1065

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	0.4549	0.22745	0.050467
S2	2	0.1682	0.0841	0.006161
S3	2	3.7156	1.8578	0.550201
S4	2	0.4754	0.2377	0.04067
S5	2	0.1526	0.0763	0.001824

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	4.678289	4	1.169572	9.006115	0.01657	5.192168
Within Groups	0.649321	5	0.129864			
Total	5.327611	9				

ANOVA for 24°&8mm axial height fully cemented

S1	S2	S3	S4	S5
0.1143	0.0367	0.027	0.0807	0.0173
0.0278	0.037	0.0378	0.0394	0.0699

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1	2	0.1421	0.07105	0.003741
S2	2	0.0737	0.03685	4.5E-08
S3	2	0.0648	0.0324	5.83E-05
S4	2	0.1201	0.06005	0.000853
S5	2	0.0872	0.0436	0.001383

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	0.002121	4	0.00053	0.43923	0.777229	5.192168
Within Groups	0.006036	5	0.001207			
Total	0.008157	9				

The dynamic loading data statistical analysis

S1 Dry	S2 Dry	S1 W	S2 W	S1 W&A	S2 W&A	S1 A	S2 A
-3.7	54	17	-110	-77	-17	-97	-8.5
20	-44	56	1.4	-35	-19	-18	-75

Anova: Single Factor

SUMMARY

Groups	*Count*	*Sum*	*Average*	*Variance*
S1 Dry	2	16.3	8.15	280.845
S2 Dry	2	10	5	4802
S1 W	2	73	36.5	760.5
S2 W	2	-108.6	-54.3	6204.98
S1 W&A	2	-112	-56	882
S2 W&A	2	-36	-18	2
S1 A	2	-115	-57.5	3120.5
S2 A	2	-83.5	-41.75	2211.125

ANOVA

Source of Variation	*SS*	*df*	*MS*	*F*	*P-value*	*F crit*
Between Groups	17850.85	7	2550.121	1.117007	0.435297	3.500464
Within Groups	18263.95	8	2282.994			
Total	36114.8	15				

12.3. Published paper

Compressive stress-strain behavior of cast dental restoration in relation to luting cement distribution

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Compressive stress-strain behaviour of cast dental restorations in relation to luting cement distribution

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Abstract. This work is concerned with the performance, under dynamic compression, of the adhesives used to simulate cementation of gold crowns onto nickel chromium dies. A measurement system, based on the mounting of strain gauges onto the outer surface of the crowns, has been developed, which allows a semi-quantitative estimate of the state of adhesion. A preliminary study was carried out where miniature gauges were bonded onto the buccal, lingual, mesial and distal surfaces of gold crowns cast to fit onto a chromium-cobalt die, as a precursor to the final design of the instrumented crown. The crowns were then loaded in compression periodically measuring the strain at all four gauges.

The results showed the load-strain relationship in fully, partially and uncemented crowns, along with repeated load-unload behaviour to close to the ultimate strength of the bond. The load-strain relationship is linear and repeatable and the slope varies over a factor of about 30 between the un-cemented and fully cemented crown, demonstrating that there is significant sensitivity to bond integrity. These results were used to determine the parameters for a systematic study of S-N relationships in restorative dentistry crowns.

Keywords: Restorative dentistry, adhesives, mechanical testing

1. Introduction

In restorative dentistry, it is common to use dental stone as a die for metallic crown whose internal surface is investment cast as a close fit to the die (Fig. 1). It is also common to use a die spacer on the die to control the gap between crown and die and hence the deposition of the luting cement. This work relates to the integrity of the bond between the die and the crown as revealed by the response, measured as strain on the crown wall, under compressive loading with a variety of simulated bond conditions.

For cementation of a full cast crown, a physiological force is generated over the preparation during crown seating (Fig. 2). This force generates a pressure due to the close proximity of the fitting surface of the crown to the preparation surface as the crown comes close to the finish line of the preparation. Cement is normally expressed from the base, but it is clear that the rheological properties of the cement and the nature of the gap between the two components will influence the distribution of the cement over the available gap and hence the integrity of the bond.

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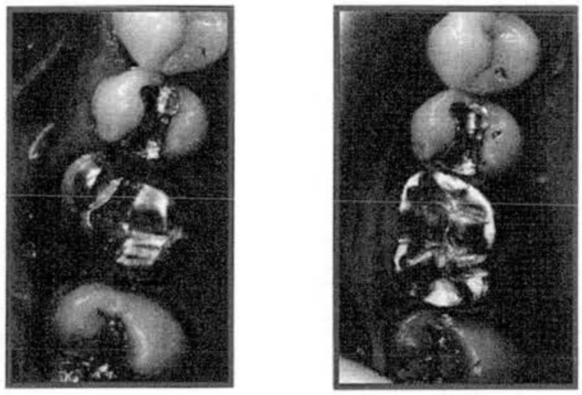


Fig. 1. Typical full cast crown; left milled crown restored with amalgam, right Gold crown in place.

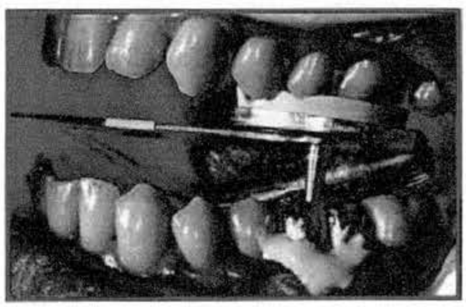


Fig. 2. Typical cementation process for full cast crown.

It is known that the degree of adaptation of the crown to the preparation surface plays an important role in the strain distribution on the axial surface of the crown and that the distribution of the luting cement affects this strain distribution. There have been many attempts to improve crown seating using available luting cements including; the use of internal surface relief [6], occlusal surface venting [9], using an escape way on the proximal surface on the preparation not duplicated on the fitting surface of the crown [15], and the application of vibration and pressure at the beginning of cementation [12]. The basis of all of these attempts was to decrease the maximum hydraulic pressure during crown seating, although each has a limited degree of mitigation. A truncated crown has also been used by some others for experimental purposes [8]. In this study, the crown and the die used were different from those used in clinical situation in their shape, mechanical property, and in the quality and location of loading.

The current work is aimed at understanding the cement distribution between crown and tooth preparation and to examine how, and if, this changes under static and dynamic compressive loading. To meet these aims, it has been necessary to develop a means of measuring the efficacy of the luting cement without sectioning the assembly, and it is this method that we present here.

2. Materials and methods

In order to develop the measurement approach, an idealised system was adopted, consisting of truncated conical dies, made from a nickel-chromium alloy with corresponding gold crowns cemented onto the die using conventional zinc phosphate cement. Strain gauges were installed onto the axial surface of the crown prior to compressive loading.

Metal dies were used to avoid fracture or distortion of the die during testing. Specifically a nickel chromium alloy was used because of its hardness and abrasion resistance under repeated cementation and crown removal, Chan [2], Lockwood [11] and Cassidy [1]. Wax was applied directly onto the metal dies, because it has been shown that the degree of adaptation or misfit of a crown that has been waxed onto a metal die is less than for a crown waxed on a stone die, Fusayama et al. [5]. Also, Eden et al. [4] have shown that base metal alloys have inferior fit on metal dies, and so a noble metal crown was chosen.

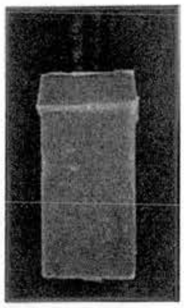
The dip technique was used for waxing of the gold crown, because it has been shown to give a reproducible crown thickness of about 0.5 mm, Kovarik et al. [10]. The control of crown thickness was seen to be important following Sugita et al. [14], who emphasised that, as metal thickness increased, the deformation of the crown decreased and the cement failure load increased. They reported that an axial metal thickness of 0.3 mm gave rise to cement failure at the smallest loads although the type of casting alloy had no significant effect on cement fracture.

The main methods used in dental experimental mechanics are strain gauges, brittle lacquer, and photoelastic techniques, Hood [7]. Palamara et al. [13] reported on the importance of the size of the strain gauge relative to the surface area of the sample, and on the difficulty in attaching very small gauges to make sufficiently accurate measurements.

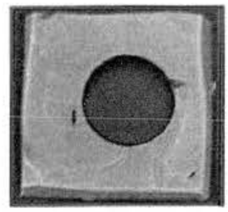
2.1. Die preparation

Figure 3 illustrates the key stages in preparation of the dies. A cylindrical plastic rod of diameter of 11 mm was used as the pattern and a plastic container fitted over it as an impression tray. An impression was then taken for the plastic rod using vinyl poly siloxane impression material (Coltene/Whaledent AG). A single mix impression technique was used where the light and the heavy bodies were mixed at the same time and the impression was left for one day. Using Type IV dental stone (Moldastone®, Heraeus Kulzer), a master die/model of the impression was cast and left to set for one day. The model was then mounted longitudinally on a moveable jig with the help of surveyor (Krupp Dental, Dentagraph) and retained by a low expansion articulating plaster (Shera, Werkstoff-Technologie). The model was milled using a Type 990 KaVo milling machine fitted with a 6° taper bur to produce a crown preparation shape with 12° total occlusal conversion (TOC), an axial wall height (from the top to the finish line) of 6 mm, and a diameter between the opposing finish lines of 10 mm. The depth of trimming was about 0.5 mm, the finish line was chamfered and the occlusal surface was flat.

Later, a second impression was taken of the milled model again using vinyl poly-siloxane impression material, and this was taken as the master model. The impression was again left for a day, removed from the tray and split longitudinally using a number 11 surgical blade. The two impression halves were reassembled and returned to the impression tray. A blue inlay hard casting wax (Kerr, USA) was melted into the impression to give 10 replicas of the impression. The wax was left to cool and solidify for about 45 minutes at which point the whole set was removed from the tray and the wax block was removed from the impression. The wax blocks were then sprued, invested using a phosphate bonded investment (Hervest® Speed E, Heraeus Kulzer) and left for about one hour to set. The investment preparation (powder to liquid ratio, 180 g of powder to 39 ml liquid per die according to the manufacturer's instructions for non-precious alloys) was stirred by hand using a spatula to wet the powder and then put into a vacuum mixer for 90 seconds. The mixed investment was poured into the casting ring whilst being vibrated. The set investment was put in an oven at 900°C for about an hour and then cast in a nickel chromium alloy



Impression tray on plastic rod



Vertical view of impression

Impression poured in type IV stone & mounted using dental surveyor

Impression in dental stone on movable jig

Master model preparation in milling machine

Nickel Chromium cast model with die spacer

Fig. 3. Key stages in preparation of dies.

(Heraeus Kulzer). The castings were left to cool on the bench prior to being divested and then blasted using 50 μm aluminium oxide powder (Fino). The sprue base was cut off with a carborundum disc and finally polished.

The bases of the ten dies were cut with a tungsten carbide tip in a centre lathe to make the base perpendicular to the long axis of the die, and then a hole was drilled up through the centre of the base using a milling machine with a 2.5 mm diameter cobalt drill. The hole was then tapped and a stainless steel screw was fitted into the tapped hole. The idea of the central screw was to permit removal of the crown after testing with minimal risk of damage to the die, crown or strain gauge. For totally cemented crowns, the screw had to be turned to flush with the flat occlusal surface of the metallic die in order to prevent any keying of the cement which might influence the stress-strain behaviour on the cement lute.

2.2. Crown fabrication

For the main set of measurements, two layers of die spacer (Tru-Fit, Geo.Taub inc. Jersy City,USA) were applied to the metallic dies to within 1 mm of the margins and allowed to dry for 10 minutes. A dummy application was used to determine the film thickness and this was found to be 25 μm. Separating medium (Microfilm, Kerr) was applied to the surface of the spacer and allowed to dry for 5 minutes and a wax pattern fabricated using dip technique where the nickel chromium die dipped into molten wax (Bellwax^{TM}, Kerr) for about 5–7 seconds to obtain an even thickness for the patterns of about 0.5 mm and an overall height of about 8 mm. The occlusal surface of the pattern was curved manually and checked using a wax gauge calliper (Iwanson Type Measuring, Gauge For wax) until about 0.5 mm of thickness was reached.

The margins of the wax pattern were re-melted and reinforced using hard blue inlay wax to ensure good conformity of the margins to the chamfer finish line. The wax patterns were sprued, and again invested according to the manufacturer's instructions for one hour using Hervest® Speed E, (Heraeus Kulzer) preparation. Once prepared, the investments were cast in Type IV (Ag 17%, Cu 19%, Pd 3%, Au 60%, Others 1%) yellow casting gold alloy (Bodent 60), and the cast gold crown was quenched in cold water immediately, removed from the investment and sandblasted with 50 μ aluminium oxide (Fino). The gold crowns were finished, polished and then fitted onto their metallic dies. The gold thickness was checked again at four random locations axially and occlusally using the gauge calliper and found to be close to 0.5 mm all over. Ten crowns were produced, one for each die. The fit of each crown was visually verified under an optical microscope at 4 $\times$ magnification. The crowns and the dies were scribed to assist fitting and avoid rotation.

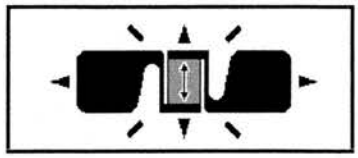
2.3. Strain gauge installation

Two miniature linear strain gauges (EA-06-031EC-350, Measurements Group) were bonded (M-Bond 200, Measurements Group) opposite each other on the surface of each gold crown about 1 mm above the finish line. The dimensions of the gauges are shown in Fig. 4.

The gauges were wired using a three-wire circuit for a single active gauge (quarter bridge) and hence each gauge gave an independent measure of strain on the axial surface. Each gauge was amplified independently, and the amplifiers were calibrated on an Aluminium-Copper alloy strip of known elastic modulus (72.4 GPa), care being taken to ensure that the strain remains below yield (approximately 2000 microstrain) and measurements being taken during increasing and decreasing load. The calibration curves for the two amplifiers used are shown in Fig. 5, and it can be seen that the measurements on increasing and decreasing load (shown as different symbol types) are highly reproducible with little or no hysteresis.

2.4. Cementation procedure

The cement used was a zinc phosphate (De Trey®Zinc, Dentsply Ltd.) and was applied to ISO specification 9917:1991 for testing of dentistry-water-based cements, where the powder: liquid ratio is 2.8g:1g. The cement was mixed with a spatula on a glass slab which had been refrigerated at 5 °C, and the room temperature during mixing was 19 $\pm$ 2°C. The powder to liquid ratio was according the manufacturer's recommendations and the mixing was carried out according to the recommendations of Eames et al. [3] where the powder was divided into equal increments and mixed with the liquid over half the area of the glass slab. Each increment of powder was incorporated into the liquid at 15 second



Description	Measurement
Gauge length	0.79 mm
Overall length	1.07 mm
Grid width	0.81mm
Overall width	3.56 mm
Matrix length	4.3 mm
Matrix width	5.8 mm

Fig. 4. Strain gauge EA-06-031EC-350 (from Measurements Group Catalogue).

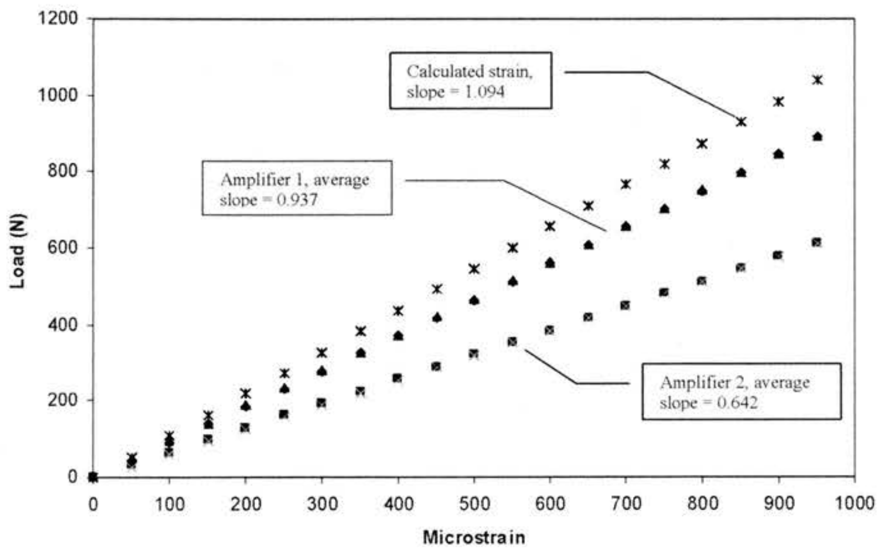


Fig. 5. Strain gauge amplifier calibration on aluminium alloy strip.

intervals and the time from the addition of the first increment until completion of the mix was $1\frac{1}{2}$ minutes.

Partial cementation was achieved by brushing the cement onto the axial surface of the die avoiding covering the occlusal surface. For total cementation, the crown was half-filled and the periphery was painted with the mixing spatula to make sure that the entire area of the fitting surface of the crown was covered. The mounting screw was put into the die and made flush with the occlusal surface. The

Test assembly

Cemented sample under test

Partially cemented crown after removal from die

Fig. 6. Crown-die assembly and testing.

direction of fitting the strain-gauged crown onto the die was kept consistent using the scribed marks mentioned above. The cemented crown-die assembly was first pressed manually and subjected to a static force of 5 kg for 2 minutes using a cementation jig.

Following completion of experiments, the uncemented and partially cemented crowns were recovered in order to re-use the strain gauges. For partially cemented crowns, this was achieved by tightening the mounting screw from the base of the die, and it was found that the crown could be recovered without damage. For re-use, the adhering cement was carefully removed with a Wards curver, and it was found that this could be done without damage to the strain gauges. Figure 6 shows a completed assembly prior to testing, and a partially cemented crown which had been successfully removed after testing.

2.5. Load application and experimental procedure

All crown-die assemblies were subjected to axial compressive loading at a crosshead speed of 0.5 mm/min using a test machine (Instron Model 5567) and the two strain gauge outputs measured as a function of applied load. Tests were generally repeated five times for each die and the results averaged. Three distinct series of test were carried out, each with the objective of assessing to what extent, and how reliably, the strain gauge method could be applied to determining the effectiveness of luting cement.

In Test 1, five crown-die assemblies using a 25 μm die spacer thickness, in each of three conditions, uncemented, partially cemented and fully cemented, were loaded to 220 N to determine whether the load-strain response was sensitive to the degree of cementation. In Test 2, a similar loading regime was applied to fully cemented assemblies using 3 die spacer thicknesses; none, 40 μm and 80 μm.

3. Results and discussion

Figures 7 to 9 show the results of Test 1, plotted as the measured strain against the applied load. Four individual assemblies can be identified (C1 to C4) and the two individual gauges are identified by the amplifier used (A1 or A2). Within each of the test conditions, it is immediately apparent that the variability between individual gauges is greater than any systematic variations due to the calibration of the amplifiers. This variability could be either due to accuracy and repeatability of the gauge installation itself or to the variability cement placement within a given condition. Whatever, the cause, it is equally

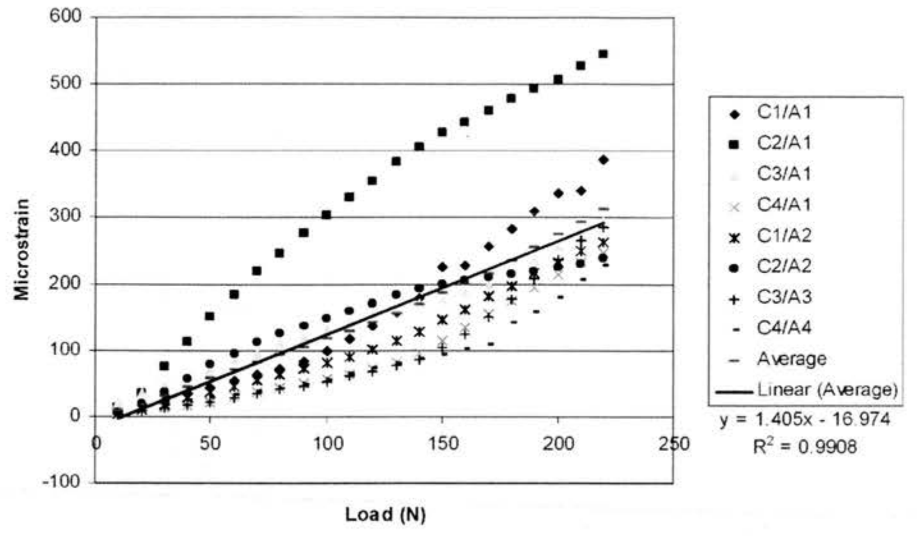


Fig. 7. Load-strain response for uncemented assemblies.

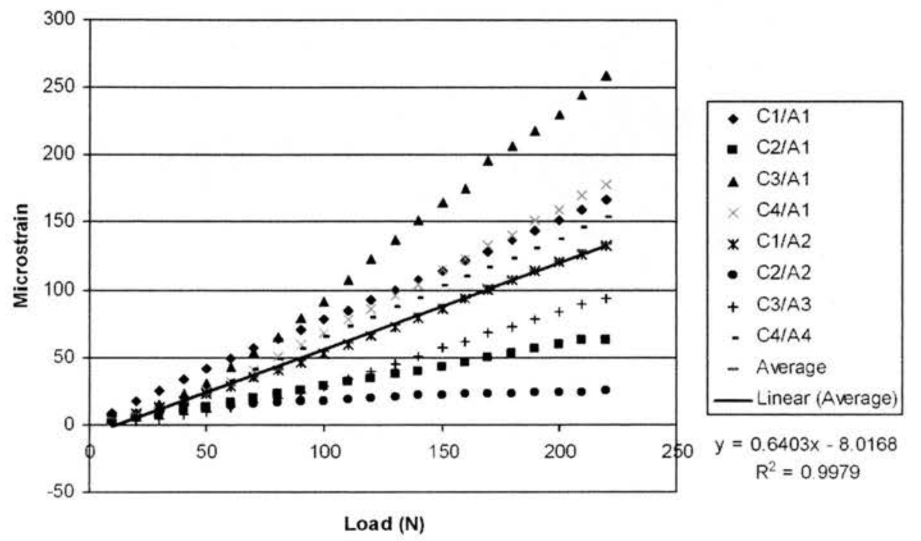


Fig. 8. Load-strain response for partially cemented assemblies.

clear that the degree of cementation has a marked effect on the strain response, with a factor of around 2 difference in sensitivity between uncemented and partially cemented assemblies, and around 20 between fully cemented and partially cemented assemblies.

Figure 10 shows the results of Test 2, this time plotted as averaged strain across all gauges on all assemblies as a function of applied load. Here it is clear that, whereas the use of no die spacer results in reduced strain response, there is very little change as the die spacer thickness is increased from 40 μm to 80 μm.

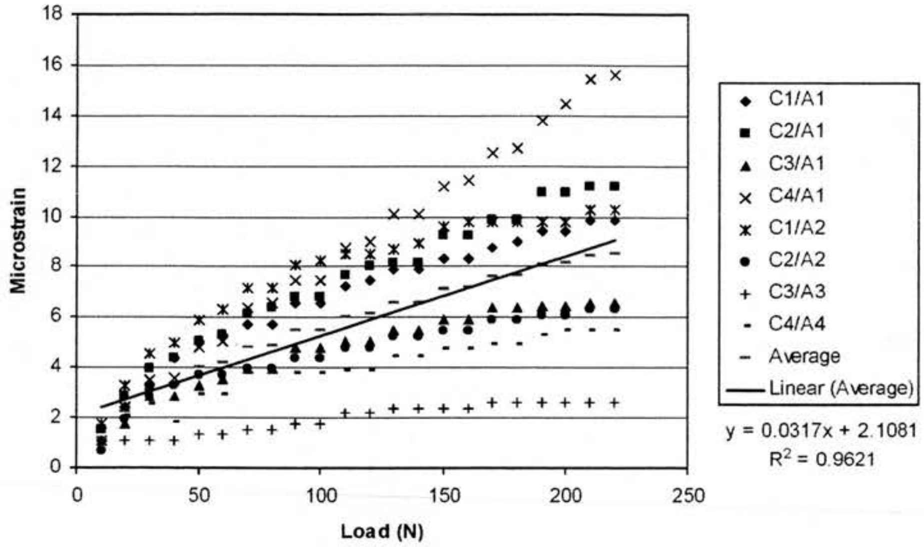


Fig. 9. Load-strain response for fully cemented assemblies.

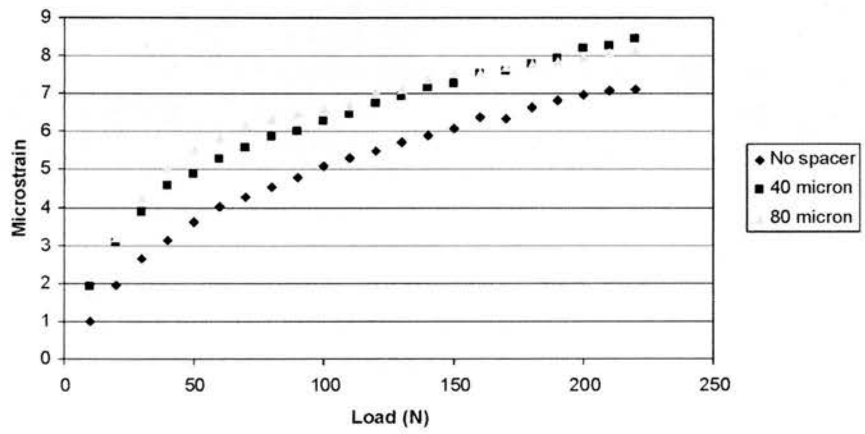


Fig. 10. Effect of die spacer thickness on load-strain response for fully cemented assemblies.

4. Conclusion

The results show that, with appropriate care in the preparation of samples and installation of strain gauges, measurements on the axis of the crown can be used to determine the condition and effectiveness of the luting cement. This is in general agreement with Yamashita et al. [16] who reported that the existence of luting cement between the bridge and the die has a significant effect on the strain distribution under load although these authors used a cantilevered loading system.

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